Stabilometric parameter analyses and optical based motion analysis

Ph.D. Thesis

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Supervisor:

Rita Kiss, DSc.

Budapest, 2019.
### Abbreviations

<table>
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<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>AP</td>
<td>anterio-posterior</td>
</tr>
<tr>
<td>AR</td>
<td>augmented reality</td>
</tr>
<tr>
<td>BMI</td>
<td>body mass index</td>
</tr>
<tr>
<td>BW</td>
<td>bandwidth</td>
</tr>
<tr>
<td>CI</td>
<td>confidence interval</td>
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<tr>
<td>CoM</td>
<td>Center of Mass (of the body)</td>
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<tr>
<td>CoP</td>
<td>Center of Pressure (of the feet)</td>
</tr>
<tr>
<td>CV</td>
<td>coefficient of variation</td>
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<tr>
<td>EC</td>
<td>eyes closed stance</td>
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<tr>
<td>EO</td>
<td>eyes open stance</td>
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<tr>
<td>FFT</td>
<td>Fast Fourier Transformation</td>
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<tr>
<td>ICC</td>
<td>intraclass correlation coefficient</td>
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<tr>
<td>IMU</td>
<td>inertial measurement unit</td>
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<tr>
<td>IR</td>
<td>infrared</td>
</tr>
<tr>
<td>LA</td>
<td>largest amplitude</td>
</tr>
<tr>
<td>LAF</td>
<td>largest amplitude frequency</td>
</tr>
<tr>
<td>LDD</td>
<td>load distribution difference between legs during static stance [%]</td>
</tr>
<tr>
<td>LED</td>
<td>light emitting diode</td>
</tr>
<tr>
<td>LMR</td>
<td>low-medium frequency band power ration</td>
</tr>
<tr>
<td>MDC</td>
<td>minimal detectable change (with level of significance set to 0.05)</td>
</tr>
<tr>
<td>MDF</td>
<td>median Frequency</td>
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<tr>
<td>MHR</td>
<td>medium-high frequency band power ration</td>
</tr>
<tr>
<td>ML</td>
<td>medio-lateral</td>
</tr>
<tr>
<td>MPF</td>
<td>mean power frequency</td>
</tr>
<tr>
<td>OMC</td>
<td>optical motion capture</td>
</tr>
<tr>
<td>RMS</td>
<td>root mean square</td>
</tr>
<tr>
<td>RMSE</td>
<td>root mean square error</td>
</tr>
<tr>
<td>ROM</td>
<td>range of motion</td>
</tr>
<tr>
<td>RPC</td>
<td>reproducibility coefficient</td>
</tr>
<tr>
<td>SD</td>
<td>standard deviation</td>
</tr>
<tr>
<td>SEM</td>
<td>standard error of measurement</td>
</tr>
<tr>
<td>SL</td>
<td>single leg stance</td>
</tr>
<tr>
<td>SPR</td>
<td>spectral power ratio</td>
</tr>
<tr>
<td>WCB</td>
<td>whole circle bearing</td>
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## 4.4 Discussion

## 3. Contribution

### 3.1 Background

### 3.2 Practical application

### 3.3 Contribution

### 3.4 Discussion

### 3.5 MAIN CONTRIBUTIONS

#### 1. Contribution

- Background
- Practical application

#### 2. Contribution

- Background
- Practical application

#### 3. Contribution

- Background
- Practical application
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1 INTRODUCTION

1.1 Motivation

The role of modern motion analysis is increasing in the fields of biomechanical research, clinical diagnostics, rehabilitation monitoring, robotics and entertainment. The motion analysis laboratory of the Department of Mechatronics Optics and Engineering Informatics in the Budapest University of Technology and Economics (Hungary) was established to participate in biomechanical studies. In order to achieve this it was important to validate the measurement systems and methods and to define measurement procedures and laboratory protocols. Main competences of the lab are balancing capability assessment and motion analysis of humans and animals. In the case of these measurements – especially in the studies on postural control – a large number of motion characterizing parameters are presented in the literature, therefore it was important to select those parameters that could be reliably used in a standardized manner in our laboratory protocols. To publish the results of motion analysis or use the laboratory to validate new measurement approaches it is important to know and publish the accuracy of the used measurement systems. These demands inspired several years of work summarized in this thesis. Further work has been carried out to develop an affordable alternative to motion capture, for which the reference system of the laboratory with validated precision was also mandatory.

1.2 Goal and structure of the dissertation

The thesis aims to present the novel methods and validation results that are the foundations of the applied solutions in balancing capability assessment and kinematic motion analysis in our motion analysis laboratory. The thesis consists of three main parts.

The first part introduces the selection method of an independent and reliable time-distance and frequency-based parameter list that can be used in the analysis of static balancing capability measured on a standing force platform.
The second part introduces a novel accuracy validation method for optical motion capture systems based on engineering surveying methods that is exemplified on the OptiTrack camera system at our laboratory. This approach is capable to observe and calibrate scaling errors of large measurement volumes that is unprecedented in the currently applied self-evaluation of these systems and the available scientifically published validation methods.

The third part introduces a new affordable measurement method for gait analysis where augmented reality technology is used to track the kinematics of the subjects. The accuracy of a proof of concept solution to this approach is validated using the OptiTrack motion capture system that is equipped in the laboratory.
2 SELECTION OF STABILOMETRY PARAMETERS

Balancing ability maintains the line of gravity of the body within the base of support. Static and dynamic balancing can be distinguished with respect to the movement or the lack of the movement. Standing balance is called postural stability [1].

2.1 Stabilometry

Postural stability is a necessary condition for retaining and recovering balance [1], which allows the body’s center of mass (CoM) to be kept close to the same location with respect to the body stance. Vertical posture and balancing during everyday activities require a constant retention of controls that require adequate and precise somatosensory, visual and vestibular senses and a sufficiently accurate response of the nervous and musculoskeletal systems. Interpretation of postural stability and measurements and analyses with different methods are fundamental tools in the exploration of balancing capability disruptions and in the assessment of treatment effectiveness, since they are simple to perform, not burdensome for the subject, and require short testing times [1].

Stabilometry is the objective method to measure the body sway during quiet standing, i.e., stance in the absence of any voluntary movements or external perturbations. According to the definition, standing balance (or postural stability) is the ability to keep the body "motionless" in a given circumstance and in a given position, i.e., to stabilize and minimize the movements of the CoM [2]. Human balance might be characterised in dynamic conditions as well [3, 4]. With the help of the inverted pendulum principle, it can be proved that during standing, the movement of the CoM can be characterized properly by the movement of the foot centre of pressure (CoP) [2]. During standing, CoP excursions are computed from the ground reaction forces, which provide an indication of postural control during quiet standing. Stabilometry is usually based on the analysis of the time variant CoP coordinates during a bipedal or single leg stance (SL) with the eyes open (EO) or closed (EC). CoP signals are usually measured on force plates or force distribution plates that are sampled at a certain frequency [5]. Force plates measure the three-dimensional components of the single equivalent force
applied to the surface and its point of application, usually called the CoP, as well as the vertical moment of force. Force distribution plates contain numerous pressure cells on the measuring surface which measure the distributed vertical force components under the foot. Form this pressure distribution data CoP is calculated as the weighted center point of the pressure distribution map. CoP can be utilized in clinical diagnostics [6] and is extensively applied in biomechanical research studies. As a stabilometry measurement is simple to perform it suits both healthy and impaired subjects (Figure 2.1).

Figure 2.1. Stabilometry measurement of an elderly person (a) with severe knee osteoarthritis and a healthy young individual (b)
Figure 2.2. Sampled instantaneous CoP data, trajectory of the traveled path of foot center of pressure and interpreted directions

From the sampled CoP data (Figure 2.2) several time-distance or frequency-based parameter can be calculated. Numerous neural or physiological alterations significantly modify postural stability and are reflected in CoP displacement. An instance of the neurological alterations mentioned is Parkinson’s disease whose early diagnosis can be reportedly predicted based on CoP analysis [7]. CoP analysis can reveal biomechanical effects of physiological alterations such as discrepancies in the postural stability of patients with different degrees of knee osteoarthritis [8] or between bilateral or unilateral involvement [9].

2.1.1 Redundancy of the CoP parameters

The movement of the CoP is influenced by many factors, such as age and gender [10], diseases [11], or measurement setups [12]. The International Society for Posture and Gait Research standardized many aspects of static stabilometry measurements in 2009 [12], but a wide variety of parameters that characterize CoP displacement is still used by researchers [11, 13]. From the two-dimensional CoP coordinates acquired during the measurement interval, many CoP parameters can be derived that can be
classified into time-distance based and frequency type measures. The load distribution difference (LDD) \(^1\) between the two lower limbs is another sensitive stabilometry parameter that was successively used for assessing differences in the postural stability of unilaterally and bilaterally involved knee osteoarthritis patients in previous research studies [14]. This parameter could also be used in the discrimination of early medial knee osteoarthritis [15].

The parameters often undergo statistical analyzes, where independency of parameters is beneficial. Many of the alternately used CoP parameters contain redundant information. Redundant parameters can be replaced by new parameters based on their combination with other parameters; for example, by the ratio of the same parameters in anteroposterior (AP) and mediolateral (ML) directions (Figure 2.2). These parameters can analyze CoP displacement from a new point of view. To assess the responsiveness of these new parameters variance analysis can be performed on groups with knowingly different postural characteristics or in case of a relatively homogeneous group in different stance types as significant intrapersonal differences also occur between stance types [16, 17]. These differences are similar to those between healthy subjects and subjects with impaired balance.

### 2.1.2 Reliability of CoP parameters

Many researchers have analyzed the test-retest reliability of certain CoP parameters on homogeneous groups of test subjects, e.g., on subjects with incomplete spinal cord injury [18], hip osteoarthritis [19], knee osteoarthritis [20], healthy elderly people [21], musculoskeletal disorders [22], stroke survivors [23], healthy young adults [10, 24-27] or healthy subjects of various ages [28]. Most studies that focused on CoP parameter reliability following the conventional test-retest analyses required that the participants perform two sessions on different days or separated by at least one-hour time intervals [19, 20, 22, 23, 25]. Within a session, there could be multiple averaged measurements. Other studies performed multiple measurements sequentially to analyze

---

\(^1\) The name and abbreviation of the studied parameters are highlighted in the text in italics
the intrasession reliability [21, 27]. Some authors established the existence of postural balance changes on different days [26] or even during the day [29, 30]. The most widely used reliability measurements by these studies include the intraclass correlation coefficient (ICC), standard error of measurement (SEM), minimal detectable change (MDC) and coefficient of variation (CV), which is also known as the relative standard deviation [28]. The results of most of these studies are well summarized in a literature review by Ruhe et al. [13], while later studies continued to research the reliability of new sophisticated stabilometry parameters [27].

Many of the above reliability studies analyzed the reliability of CoP parameters with certain measurement setups, protocols and sampling conditions, while some research studied the effects of different settings on the parameter reliability. Recommendations of stabilometry standardization are based on these studies, which changed during the years in the light of newer results. While earlier, a minimum of 60-second sampling interval was recommended [31], a more recent standardization initiative recommends a minimum of 25-second sampling interval for time-distance type parameters in stabilometry [12]. However, due to the large number of CoP parameters, not every recommendation covers all parameters. Many of these parameters contain redundant information [32], such as the path length and average path velocity in the case of the standardized measurement interval [18].

Section 2.1.1 introduces the need for a reduction on the large number of CoP parameters to select the independent CoP parameters that are sufficiently sensitive to show the differences between the different standing conditions [33, 34]. Our research resulted separate time-distance and frequency type parameter sets. The reliability of the resulting parameter list has to be analyzed.

Based on the literature review of Ruhe et al. [13], only Santos et al. [24] studied the reliability of the time and frequency domain parameters together, but only during bipedal stances. Reliability findings in a bipedal stance with the eyes opened and closed are valid not only for young, healthy persons but also for the elderly or orthopedically or neurologically altered patients [13]. Reliability analysis of CoP parameters during a single leg stance is also crucial because the single leg stance is an important test case in the analysis of the effect of the foot structure on the postural stability [35–37]; however, CoP parameter reliability on single leg stances has been rarely studied and only on
averaged values [38] in shorter (30-second) trials compared with the recommended 60-second sampling interval for some studied frequency type parameters [31].

2.1.3 Outline of the present research

The present research aims to reduce the number of the time-distance type and the frequency type CoP parameters by excluding those that contain redundant information, also replaced multiple parameters with summarizing parameters which based on the replaced parameters give a new and independent approach of the same phenomenon.

On the basis of the literature [13, 19, 20, 22–25], the second aim of the present research is to evaluate the test-retest reliability of the identified independent time-distance and frequency-based CoP parameters. The bipedal stance with the EO, EC and SL stances were involved in our measurement protocols, and the reliability of the parameters was determined for both 30- and 60-second sampling intervals with repeated standing trials of healthy young participants on a force distribution plate. It is hypothesized that by performing the correlation analysis based parameter number reduction and reliability analysis on the remaining parameters a unified parameter set can be defined that non-redundantly covers every aspect of CoP based balancing capability assessment. It is expected that the summary parameters (such as path length) will be more reliable in our parameter set owing to their averaging effect compared to extremity- and frequency-based parameters, which are more influenced by randomness.

2.1.4 Methodology of the applied stabilometry measurements

2.1.4.1 Equipment

Stabilometry was performed on a Zebris FDM-S multifunctional force distribution measuring plate (320 mm × 470 mm measuring surface with 1504 load cells) (ZEBRIS GmbH, Isny, Germany) in the motion analysis laboratory of the Department of Mechatronics, Optics and Mechanical Engineering Informatics of the Budapest University of Technology and Economics (Hungary). During the measurements, the room was silent, and no other activities were performed in the room that could distract the measurement. The room temperature was comfortable for the participant’s attire.
2.1.4.2 Measurement procedure

The measurement procedure has been registered online which includes the detailed guidelines of the procedure ([https://dx.doi.org/10.17504/protocols.io.ns6dehe](https://dx.doi.org/10.17504/protocols.io.ns6dehe)). Before performing stabilometry, the basic anthropometrical data for all participants were registered. Specifically, the body height was measured and recorded in cm; and the body mass was measured to the nearest 0.1 kg with an electronic weight scale with the participants wearing shorts and a T-shirt. During the measurements, the participants were barefoot.

The studied stance types were bipedal stance with the eyes open (EO) or eyes closed (EC) and stance on the dominant leg (SL). In each session, the order of the stance types was randomized and was continued successively without rest. Before the test, the dominant side of each participant was determined by a balance recovery test. The perturbation was a push from the tester applied to the subject at the midpoint between the scapulae from directly behind the subject and sufficient to require the participant to respond by taking a step. The leg that the subject moved to recover balance was considered to be the dominant leg [39]. The investigated subjects were positioned in bipedal standing, with the distance between the two ankle joint centers equal to the distance between the right and left anterior superior iliac spines (Figure 2.1). Both limbs were in full knee extension, the heels were aligned in a line, and the feet were parallel and faced forward, with arms resting by the sides. During the stance on the dominant leg, the dominant leg was in full knee extension, and the non-dominant leg was flexed at 90 degrees. Since a 60-second long single leg standing trial is challenging, in case the subject had to put down the other leg or showed substantial arm or whole body movements, the measurement was aborted and repeated. This event was rare because the participants were young, healthy individuals capable of performing a 60-second single leg stance. The exact foot placement was maintained during the trials and was visually controlled since its changes could affect the stabilometry parameters [40]. During stances with the eyes open (EO and SL), the eyes were focused on the wall at eye level 5 meter from the subject. The measurements started after the initial transients of the subject’s self-adjustment. The adjustment time was usually approximately 10 seconds, but at a minimum 5 second wait after the subject stood on the platform. The
pressure distribution under the feet was monitored by the measuring person on the computer screen to validate if the subject was standing still.

2.1.4.3 Ethical approval

The participants received detailed oral and written information about the risks and benefits of the study. Each participant gave signed informed consent and was given the opportunity to withdraw from the study at any time. The study was approved by the Hungarian National Science and Research Ethics Committee (114/2004).

2.2 Parameter reduction of the time-distance and frequency type CoP parameters

2.2.1 Methods in parameter reduction

The measurement procedure and the used equipment and ethical approval is detailed in 2.1.4. Each participant performed one from each detailed stance types.

2.2.1.1 Participants

25 young healthy individuals were involved in the study (22 males, 3 females: 22.34 ± 0.97 years, 74.34 ± 4.23 kg, 1.711 ± 0.085 m, BMI: 25.3 ± 2.8). As the main interest was on the within-subject effect of the different stance types by repeated measures, participants were not separated by gender, height or weight to achieve higher sample homogeneity. The participants did not suffer any musculoskeletal injury and did not go through surgery in the last ten years.

2.2.1.2 Assessed parameters

Force distribution was recorded using the Zebris WinPDMS processing software v1.2 (Zebris GmbH, Isny Germany). This software runs only on a Microsoft® Windows® XP® 32-bit operation system. The sampling frequency was 30 Hz which satisfies the Shannon sampling theory, considering that 90% of the spectral power is under 2 Hz [17] and the frequency content of the CoP signal above 5-10 Hz is considered noise as it is often low-pass filtered [41–43].

Raw force distribution data was obtained from the WinPDMS software and was processed in custom processing software written in LabVIEW 2013 (National
Instruments Inc., Austin, Texas). This software extracts instantaneous CoP coordinates from the force distribution data and calculates the frequency analysis data from it in AP and ML directions (Figure 2.2). From the CoP position signals power spectrum was obtained using Fast Fourier Transformation (FFT) with Hanning window. The acquired frequency resolution was 0.016 Hz.

The most widely used CoP parameters have been selected from the literature ([13, 19, 20, 22–25]) for the parameter study. Some newly defined CoP parameter have also been included. These are mostly AP-ML range ratios of other popular CoP parameter in the hope of earning parameters which demonstrate the standing balance in newer perspectives. The included CoP parameters are introduced in Table 2.1.

**Table 2.1.** Studied parameters

<table>
<thead>
<tr>
<th>Parameter name</th>
<th>Dimension</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Time-distance parameters</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Confidence ellipse axes</td>
<td>mm</td>
<td>The length of the minor and major axes of the 95% confidence ellipse</td>
</tr>
<tr>
<td>Confidence ellipse area (CE area)</td>
<td>mm²</td>
<td>The area of the 95% confidence ellipse around the CoP trajectory.</td>
</tr>
<tr>
<td>Confidence ellipse angle</td>
<td>degree</td>
<td>The angle of the 95% confidence ellipse from where the major axis is parallel to the AP axis. Positive direction counterclockwise.</td>
</tr>
<tr>
<td>Confidence ellipse axis ratio* (CE axis ratio)</td>
<td>-</td>
<td>The ratio between the major and minor axes of the 95% confidence ellipse that describes the shape of the CoP’s trajectory expansion.</td>
</tr>
<tr>
<td>Path length</td>
<td>mm</td>
<td>The length of the total CoP trajectory during the measurement.</td>
</tr>
<tr>
<td>Average path velocity</td>
<td>mm/s</td>
<td>The CoP path length divided by the sampling time of the standing trial.</td>
</tr>
<tr>
<td>Maximum path velocity</td>
<td>mm/s</td>
<td>The filtered maximum distance between consecutive CoP points divided by the sampling interval.</td>
</tr>
<tr>
<td>AP and ML standard deviations</td>
<td>mm</td>
<td>Standard deviation of the CoP coordinates in the AP and the ML direction.</td>
</tr>
<tr>
<td><strong>AP and ML ranges</strong></td>
<td><strong>mm</strong></td>
<td>Difference of the maximum and minimum coordinates traveled by the CoP in the AP and the ML directions</td>
</tr>
<tr>
<td>----------------------</td>
<td>-------</td>
<td>------------------------------------------------------------------------------------------------------------------</td>
</tr>
<tr>
<td>AP-ML range ratio*</td>
<td>1</td>
<td>The ratio of the largest CoP path expansions in the anteroposterior (AP) and mediolateral (ML) directions that describes the relation of the largest random errors of postural control between the two anatomical directions.</td>
</tr>
<tr>
<td>Anterior (AP+) and Posterior (AP-) maximum deviations*</td>
<td>mm</td>
<td>The maximum excursions in the anterior and posterior direction relative to the average CoP point in the AP-ML plane (Figure 2.3)</td>
</tr>
<tr>
<td>The angle of the Anterior (AP+) and Posterior (AP-) maximum deviations*</td>
<td>degree</td>
<td>The angle of the Anterior (AP+) and Posterior (AP-) maximum deviations from the AP axis. Positive direction counterclockwise (Figure 2.3)</td>
</tr>
</tbody>
</table>

**Frequency parameters**

| **Largest amplitude during balancing (LA) and largest amplitude frequency (LAF)*** | **mm and 1/sec** | The largest continuous motion in both the AP (Figure 2.4) and the ML directions, which are not necessarily equal to the corresponding CoP range. This parameter is similar to the sub-movement size that was defined by [44] for targeted CoP movements. Peaks and valleys of the signal are located and the largest peak-valley or valley-peak transition in amplitude is considered the double of the amplitude of the modeled sine function. The temporal difference between the selected signal peak and valley is the half period of the sine function. From these variables of the modeled sine function the LA and LAF are calculated. |
| **Frequency power ratios between low-medium and medium-high frequency bands (LMR, MHR)** | 1 | Provide information about the power distribution of the postural sway in the frequency domain. The defined limits of the compared frequency bands are low- (0-0.3 Hz) medium- (0.3-1 Hz) and high frequency (1-5 Hz) bands [16]. |
| **Median frequency (MDF) in AP and** | Hz | The frequency value which separates the power spectrum into two equal energy areas. MDF was calculated as proposed by Oskoei and Hu [45] |
ML direction according to the following equation:
\[
\sum_{j=1}^{\text{MDF}} P_j = \sum_{j=1}^{M} P_j = \frac{1}{2} \sum_{j=1}^{M} P_j
\]
where \( P \) is the power of each frequency bands.

<table>
<thead>
<tr>
<th>Spectral power ratio (SPR)*</th>
<th>1</th>
<th>The ratio of the total spectral power in the AP direction and the total spectral power in the ML direction. SPR characterizes the rate of the power distribution of the postural sway frequencies in the AP/ML directions.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bandwidth (BW) in AP and ML direction</td>
<td>Hz</td>
<td>The width of the frequency band in Hz in which 99% of the total power of the CoP signal is located.</td>
</tr>
<tr>
<td>Mean power frequency (MPF)</td>
<td>Hz</td>
<td>A weighted average frequency where the ( f_j ) frequency components are weighted by their ( P_j ) power. ( M ) is the number of frequency bins. MPF is calculated as proposed by Oskoei and Hu [45], according to the following equation: ( \text{MPF} = \frac{\sum_{j=1}^{M} f_j P_j}{\sum_{j=1}^{M} P_j} )</td>
</tr>
</tbody>
</table>

**Other**

| Load distribution difference (LDD) | % | Shows the difference in the weight load on the lower limbs. This parameter is not derived from CoP motion, but it is used by the original Zebris WinPDMS software together with the CoP parameters and is proven to be very useful in biomechanical analyses [14, 15]. |

*indices those mostly ratio type parameters which has been newly introduced in the study

![Figure 2.3. CoP AP+ (\(d_1\)) and AP- maximum deviation (\(d_2\)) and angles (\(\alpha, \beta\))](image-url)
Figure 2.4. CoP time domain determination of the largest amplitude and the corresponding frequency

2.2.1.3 Statistical analysis

The usability of the newly defined CoP parameters (*CE axis ratio, AP-ML range ratio, AP+, AP- deviations and angles, LA, LAF, SPR*) was examined using statistical analysis. To define which set of parameters can be replaced by fewer parameters correlation analysis was used on the resulting parameters of summed bipedal stance trials.

The behavior of the newly defined parameters in respect to the different stance trials was analyzed by a multivariate ANOVA. The alteration of postural stability thus CoP movement between different stance types (EO, EC, SL) have already been reported many times, therefore it is expected that the studied parameters indicate differences between the three stance types within subjects. The ANOVA requires meeting the following assumptions.

- Continuity of the dependent variables which in our case is guaranteed by their nature.
- The independent variable must have at least 2 groups. In our study three groups for the three kinds of trials are given.
- Outliers and extremes should be excluded from the analysis to avoid distortion and gain higher statistical power for the analysis.
• Distribution of the dependent variable in the related groups should be approximately normally distributed.

• The variances of the differences between all combinations of related groups must be equal, also known as sphericity.

The correlation matrix was also utilized to ensure the independency of the variables by excluding variables that are strongly correlated and replacing some with the newly defined parameters which’s independency was also verified. By excluding parameters that are strongly correlated to others, reduction in the parameter set can be achieved while keeping the diverse information gain from the different calculated parameters. Strong correlation was considered above 0.8 as suggested by Chan in [46], but moderate correlations above 0.5 were also taken into account.

Extremes and most outliers were manually removed from the dataset based on box plots considerably decreasing our sample size from 32 to 25, but still improving the statistical power thus reliability of the analysis. Several parameters in one participant’s single leg trial were extremes, because he failed the trial and had to put down his leg, therefore he was also excluded from the study. Normal distribution of the dependent variables was analyzed by Shapiro-Wilk test of normality, which is more appropriate for sample size below 50 unlike the Kolmogorov-Smirnov test of normality. Not normally distributed variables can be transformed to normal distribution, but this is not necessarily required because ANOVA is quite robust for the violation of this assumption [47]. Sphericity was tested with Mauchly’s test of sphericity. For the variables that do not meet the required sphericity Greenhouse-Geisser correction was applied when evaluating significance level of the within-subject effects. Overall significant difference between the means of the three stance types was calculated for each variable. To see where the specific differences occurred pairwise comparisons were performed between groups for each variable. During the statistical analysis the level of significance is $\alpha =0.05$. 
2.2.2 Results of CoP parameter reduction

2.2.2.1 Redundancy of time-distance type CoP parameters

The results of the correlation analysis of time-distance type CoP parameters are shown in Table 2.2. The 95% confidence ellipse area has naturally strong correlation with the length of its minor (0.896) and major axis (0.86). The area also has a strong correlation with the standard deviations in AP (0.769) and ML (0.821) directions and trajectory ranges in the same directions (AP: 0.686, ML: 0.788). There are strong correlations between the standard deviations and ranges of the CoP in both AP (0.887) and ML directions (0.917). The confidence ellipse axes also correlate with the standard deviations and ranges (Minor axis – ML SD: 0.885, Minor axis – ML range: 0.896, Major axis – AP SD: 0.937, Major axis – AP range: 0.827). The average CoP velocity is in a linear connection with the path length (correlation = 1.0). There is no significant correlation between the newly defined parameters (Axis ratio, AP-ML range ratio, Max. deviation in ±AP directions and its angle) and the previously used parameters (correlation < 0.8). However, Axis ratio and AP-ML range ratio show some redundancy (correlation = 0.744).

Based on the correlation analysis results we excluded the redundant variables; therefore we applied the variance analysis only on the following parameters:

- 95% confidence ellipse axis ratio,
- 95% confidence ellipse area,
- path length,
- maximum velocity,
- AP-ML range ratio,
- AP+ deviation,
- AP- deviation,
- AP+ deviation angle,
- AP- deviation angle,
- 95% confidence ellipse angle.
### Table 2.2. Correlation matrix of time-distance based CoP parameters

<table>
<thead>
<tr>
<th></th>
<th>Minor Axis</th>
<th>Major Axis</th>
<th>Area</th>
<th>Angle</th>
<th>Axis Ratio</th>
<th>Path length</th>
<th>Average velocity</th>
<th>Maximum velocity</th>
<th>AP standard deviation</th>
<th>ML standard deviation</th>
<th>AP range</th>
<th>ML range</th>
<th>AP+ ML range ratio</th>
<th>AP+ Maximum deviation</th>
<th>AP+ Maximum deviation angle</th>
<th>AP- Maximum deviation</th>
<th>AP- Maximum deviation angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minor Axis</td>
<td>1</td>
<td>0.609</td>
<td>0.896</td>
<td>0.009</td>
<td>-0.594</td>
<td>-0.013</td>
<td>0.111</td>
<td>0.498</td>
<td>0.885</td>
<td>0.48</td>
<td>0.896</td>
<td>-0.546</td>
<td>0.651</td>
<td>0.178</td>
<td>0.518</td>
<td>-0.227</td>
<td></td>
</tr>
<tr>
<td>Major Axis</td>
<td>0.609</td>
<td>1</td>
<td>0.86</td>
<td>0</td>
<td>0.189</td>
<td>-0.055</td>
<td>0.141</td>
<td><strong>0.937</strong></td>
<td>0.614</td>
<td>0.827</td>
<td>0.579</td>
<td>0.023</td>
<td>0.664</td>
<td>0.028</td>
<td>0.761</td>
<td>0.138</td>
<td></td>
</tr>
<tr>
<td>Area</td>
<td>0.896</td>
<td>0.86</td>
<td>1</td>
<td>0.009</td>
<td>-0.259</td>
<td>-0.073</td>
<td>0.151</td>
<td>0.769</td>
<td><strong>0.821</strong></td>
<td>0.686</td>
<td>0.788</td>
<td>-0.288</td>
<td>0.692</td>
<td>0.087</td>
<td>0.681</td>
<td>-0.077</td>
<td></td>
</tr>
<tr>
<td>Angle</td>
<td>0.009</td>
<td>0</td>
<td>0.009</td>
<td>1</td>
<td>-0.052</td>
<td>0.054</td>
<td>0.129</td>
<td>-0.022</td>
<td>0.034</td>
<td>-0.022</td>
<td>0.021</td>
<td>-0.091</td>
<td>0.166</td>
<td>0.223</td>
<td>-0.168</td>
<td>0.177</td>
<td></td>
</tr>
<tr>
<td>Axis Ratio</td>
<td>-0.594</td>
<td>0.189</td>
<td>-0.259</td>
<td>-0.052</td>
<td>1</td>
<td>-0.088</td>
<td>-0.088</td>
<td>0.036</td>
<td>0.263</td>
<td>-0.471</td>
<td>0.186</td>
<td>-0.482</td>
<td>0.744</td>
<td>-0.098</td>
<td>-0.192</td>
<td>0.079</td>
<td></td>
</tr>
<tr>
<td>Path length</td>
<td>-0.013</td>
<td>-0.055</td>
<td>-0.073</td>
<td>0.054</td>
<td>-0.088</td>
<td>1</td>
<td><strong>1</strong></td>
<td>-0.019</td>
<td>-0.009</td>
<td>-0.064</td>
<td>-0.004</td>
<td>0.021</td>
<td>-0.053</td>
<td>0.059</td>
<td>0.157</td>
<td>-0.047</td>
<td></td>
</tr>
<tr>
<td>Average velocity</td>
<td>-0.013</td>
<td>-0.055</td>
<td>-0.073</td>
<td>0.054</td>
<td>-0.088</td>
<td>1</td>
<td><strong>1</strong></td>
<td>-0.019</td>
<td>-0.009</td>
<td>-0.064</td>
<td>-0.004</td>
<td>0.021</td>
<td>-0.053</td>
<td>0.059</td>
<td>0.157</td>
<td>-0.047</td>
<td></td>
</tr>
<tr>
<td>Maximum velocity</td>
<td>0.111</td>
<td>0.141</td>
<td>0.151</td>
<td>0.129</td>
<td>0.036</td>
<td>-0.019</td>
<td>-0.019</td>
<td>1</td>
<td>0.111</td>
<td>0.126</td>
<td>0.209</td>
<td>0.17</td>
<td>0.082</td>
<td>0.325</td>
<td>-0.078</td>
<td>0.044</td>
<td></td>
</tr>
<tr>
<td>AP standard deviation</td>
<td>0.498</td>
<td><strong>0.937</strong></td>
<td>0.769</td>
<td>-0.022</td>
<td>0.263</td>
<td>-0.009</td>
<td>-0.009</td>
<td>0.111</td>
<td><strong>1</strong></td>
<td>0.366</td>
<td><strong>0.887</strong></td>
<td>0.393</td>
<td>0.212</td>
<td>0.608</td>
<td>-0.048</td>
<td>0.746</td>
<td></td>
</tr>
<tr>
<td>ML standard deviation</td>
<td><strong>0.885</strong></td>
<td>0.614</td>
<td><strong>0.821</strong></td>
<td>0.034</td>
<td>-0.471</td>
<td>-0.064</td>
<td>-0.064</td>
<td>0.126</td>
<td>0.366</td>
<td><strong>1</strong></td>
<td>0.347</td>
<td><strong>0.917</strong></td>
<td>-0.62</td>
<td>0.595</td>
<td>0.236</td>
<td>-0.457</td>
<td></td>
</tr>
<tr>
<td>AP range</td>
<td>0.48</td>
<td><strong>0.827</strong></td>
<td>0.689</td>
<td>-0.022</td>
<td>0.186</td>
<td>-0.004</td>
<td>-0.004</td>
<td>0.209</td>
<td><strong>0.887</strong></td>
<td>0.347</td>
<td>1</td>
<td>0.382</td>
<td>0.311</td>
<td>0.705</td>
<td>-0.051</td>
<td>0.792</td>
<td></td>
</tr>
<tr>
<td>ML range</td>
<td>0.896</td>
<td>0.579</td>
<td>0.728</td>
<td>0.021</td>
<td>-0.482</td>
<td>0.021</td>
<td>0.021</td>
<td>0.17</td>
<td>0.393</td>
<td><strong>0.917</strong></td>
<td>0.382</td>
<td><strong>1</strong></td>
<td>-0.672</td>
<td>0.645</td>
<td>0.275</td>
<td>0.466</td>
<td></td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td>-0.546</td>
<td>0.023</td>
<td>-0.288</td>
<td>-0.091</td>
<td>0.744</td>
<td>-0.053</td>
<td>-0.053</td>
<td>0.082</td>
<td>0.212</td>
<td>-0.62</td>
<td>0.311</td>
<td>-0.672</td>
<td>1</td>
<td>-0.081</td>
<td>-0.289</td>
<td>0.113</td>
<td></td>
</tr>
<tr>
<td>AP+ Maximum deviation</td>
<td>0.651</td>
<td>0.664</td>
<td>0.692</td>
<td>0.166</td>
<td>-0.098</td>
<td>0.059</td>
<td>0.059</td>
<td>0.325</td>
<td>0.608</td>
<td>0.595</td>
<td>0.705</td>
<td>0.645</td>
<td>-0.081</td>
<td>1</td>
<td>0.142</td>
<td>0.35</td>
<td></td>
</tr>
<tr>
<td>AP+ Maximum deviation angle</td>
<td>0.178</td>
<td>0.028</td>
<td>0.087</td>
<td>0.223</td>
<td>-0.192</td>
<td>0.157</td>
<td>0.157</td>
<td>-0.078</td>
<td>-0.048</td>
<td>0.236</td>
<td>-0.051</td>
<td>0.275</td>
<td>-0.289</td>
<td>0.142</td>
<td>1</td>
<td>0.109</td>
<td></td>
</tr>
<tr>
<td>AP- Maximum deviation</td>
<td>0.518</td>
<td>0.761</td>
<td>0.681</td>
<td>-0.168</td>
<td>0.079</td>
<td>-0.047</td>
<td>0.044</td>
<td>0.746</td>
<td>0.457</td>
<td>0.792</td>
<td>0.466</td>
<td>0.113</td>
<td>0.35</td>
<td>0.109</td>
<td>1</td>
<td>0.053</td>
<td></td>
</tr>
<tr>
<td>AP- Maximum deviation angle</td>
<td>-0.227</td>
<td>0.138</td>
<td>-0.077</td>
<td>0.177</td>
<td>0.308</td>
<td>0.188</td>
<td>-0.107</td>
<td>0.265</td>
<td>-0.279</td>
<td>0.267</td>
<td>-0.3</td>
<td>0.4</td>
<td>0.014</td>
<td>-0.125</td>
<td>0.053</td>
<td>1</td>
<td></td>
</tr>
</tbody>
</table>

Bold: $|\text{correlation}| \geq 0.8$, Italic: $0.8 > |\text{correlation}| \geq 0.5$
2.2.2.2 Variance analysis of the time-distance type CoP parameters

Since the ANOVA requires independent variables among the strongly correlated variables the confidence ellipse area was kept, while the ellipse axes, CoP ranges and standard deviations were excluded. Additionally axis ratio and AP-ML range ratio was used which are uncorrelated to the other variables. Instead of the average CoP velocity maximum velocity was used. Detailed results of the variance analysis are presented in Table 2.3 and Table 2.4.

**Table 2.3.** Univariate test results (difference significance levels among the three trials for each parameter)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Significance</th>
<th>Observed power</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axis ratio</td>
<td>&lt;0.001</td>
<td>0.975</td>
</tr>
<tr>
<td>95% confidence ellipse area</td>
<td>&lt;0.001</td>
<td>1.000</td>
</tr>
<tr>
<td>Path length</td>
<td>&lt;0.001</td>
<td>1.000</td>
</tr>
<tr>
<td>Maximum velocity</td>
<td>&lt;0.001</td>
<td>1.000</td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td>0.001</td>
<td>0.960</td>
</tr>
<tr>
<td>AP+ deviation</td>
<td>&lt;0.001</td>
<td>1.000</td>
</tr>
<tr>
<td>AP- deviation</td>
<td>&lt;0.001</td>
<td>1.000</td>
</tr>
<tr>
<td>AP+ deviation angle</td>
<td>0.425</td>
<td>0.192</td>
</tr>
<tr>
<td>AP- deviation angle</td>
<td>0.112</td>
<td>0.444</td>
</tr>
<tr>
<td>Confidence ellipse angle</td>
<td>0.162</td>
<td>0.331</td>
</tr>
</tbody>
</table>

AP: anteroposterior, ML: mediolateral

**Table 2.4.** Pairwise comparison on time-distance CoP parameters between stance types

<table>
<thead>
<tr>
<th>Measure</th>
<th>Mean Difference</th>
<th>Std. Error</th>
<th>Sig.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axis ratio</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>-0.170</td>
<td>0.222</td>
<td>1.000</td>
</tr>
<tr>
<td>EO SL</td>
<td>0.606</td>
<td>0.171</td>
<td><strong>0.004</strong></td>
</tr>
<tr>
<td>EC SL</td>
<td>0.775</td>
<td>0.162</td>
<td><strong>&lt;0.001</strong></td>
</tr>
<tr>
<td>95% confidence ellipse area</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>7.352</td>
<td>4.996</td>
<td>0.458</td>
</tr>
<tr>
<td>EO SL</td>
<td>-150.926</td>
<td>10.968</td>
<td><strong>&lt;0.001</strong></td>
</tr>
<tr>
<td>EC SL</td>
<td>-158.278</td>
<td>11.688</td>
<td><strong>&lt;0.001</strong></td>
</tr>
<tr>
<td>Path length</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>-122.919</td>
<td>27.311</td>
<td><strong>&lt;0.001</strong></td>
</tr>
<tr>
<td></td>
<td>EO</td>
<td>SL</td>
<td></td>
</tr>
<tr>
<td>--------------------------</td>
<td>------</td>
<td>------</td>
<td>-------</td>
</tr>
<tr>
<td></td>
<td>EC</td>
<td>SL</td>
<td></td>
</tr>
<tr>
<td>Maximum velocity</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>-5.667</td>
<td>15.307</td>
<td>1.000</td>
</tr>
<tr>
<td>EO SL</td>
<td>-236.304</td>
<td>23.327</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>EC SL</td>
<td>-230.637</td>
<td>24.792</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>-0.376</td>
<td>0.148</td>
<td>0.051</td>
</tr>
<tr>
<td>EO SL</td>
<td>0.162</td>
<td>0.111</td>
<td>0.467</td>
</tr>
<tr>
<td>EC SL</td>
<td>0.537</td>
<td>0.137</td>
<td></td>
</tr>
<tr>
<td>AP+ maximum deviation</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>-1.054</td>
<td>0.984</td>
<td>0.881</td>
</tr>
<tr>
<td>EO SL</td>
<td>-10.504</td>
<td>1.250</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>EC SL</td>
<td>-9.451</td>
<td>1.192</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>AP- maximum deviation</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EO EC</td>
<td>0.289</td>
<td>1.115</td>
<td>1.000</td>
</tr>
<tr>
<td>EO SL</td>
<td>-9.699</td>
<td>1.282</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>EC SL</td>
<td>-9.988</td>
<td>1.465</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>AP+ maximum deviation angle</td>
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</tr>
<tr>
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<td>7.970</td>
<td>0.687</td>
</tr>
<tr>
<td>EO SL</td>
<td>8.983</td>
<td>8.816</td>
<td>0.952</td>
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<tr>
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</tr>
<tr>
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<td>1.000</td>
</tr>
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<td>EC SL</td>
<td>12.896</td>
<td>5.928</td>
<td>0.116</td>
</tr>
</tbody>
</table>

Bold: significant difference (p≤0.05)

Among the used variables described above the Shapiro-Wilk test of normality gave significant difference from normal distribution to axis ratio (0.024), maximum velocity (0.001) and AP-ML range ratio (0.024) for eyes open bipedal stance, ellipse area (0.017), path length (0.024) and maximum velocity (0.003) in eyes closed bipedal stance, furthermore axis ratio (0.018), ellipse angle (0.01) and path length (0.009) in eyes open single leg stance. The variance analysis is robust enough therefore these variables were not transformed to normal distribution. Mauchly’s sphericity test
indicated the violation of sphericity among stance types for ellipse area (<0.001), path length (<0.001), maximum velocity (0.022) and ellipse angle (0.008). For these variables the Greenhouse-Geisser correction was applied when evaluating differences among groups.

The within subjects multivariate test showed overall significant differences between the three stance types (p<0.001, Observed power = 1.000). The univariate tests show the significant differences for means between stance types (Table 2.3). Basically, most variables show significant differences between the three stance types except for the three angle type variables. Pairwise comparisons reveal among which stance types occurred the significant difference. The significance levels for pairwise comparisons are displayed in Table 2.4.

2.2.2.3 Redundancy of the frequency type CoP parameters

The results of correlation analysis are summarized in Table 2.5. The strongest correlations can be observed in AP between the mean power frequency (MPF) and median frequency (MDF) (0.846). Furthermore strong correlation can be observed between MPF and bandwidth (BW) in both ML (0.812) and AP direction (0.767). It is interesting to note that there are several moderate negative correlations. In AP direction MPF negatively correlates with LMR (-0.613). Most correlations with other parameters are produced by BW. BW in AP direction (BW-AP) (besides MPF-AP) is moderately correlated with LAF-AP (0.504) and LA-ML (0.605). In AP direction BW negatively correlates with low-medium frequency range power ratio (LMR) (-0.647) and medium-high frequency range power ratio (MHR) (-0.65). BW in AP direction also negatively correlates with MHR in ML direction (-0.603). In ML direction BW also negatively correlates with LMR (-0.557) and MHR (-0.583) and positively correlates with MDF (0.528). BWs in AP and in ML direction are also moderately correlated with each other (0.514). MHR correlates between AP and ML directions (0.642).
Table 2.5. Correlations of frequency type CoP parameters

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<td>-0.199</td>
<td>0.009</td>
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<td>0.223</td>
<td>-0.006</td>
<td>0.033</td>
<td>-0.176</td>
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<td>-0.286</td>
<td>-0.377</td>
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<td>-0.216</td>
<td>0.223</td>
<td>0.321</td>
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<td>0.504</td>
<td>0.104</td>
<td>0.484</td>
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<tr>
<td>LA-ML</td>
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<td>1</td>
<td>0.196</td>
<td>-0.345</td>
<td>-0.372</td>
<td>-0.072</td>
<td>-0.306</td>
<td>0.286</td>
<td>0.057</td>
<td>-0.479</td>
<td>0.605</td>
<td>0.118</td>
<td>0.282</td>
<td>-0.221</td>
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<td>-0.387</td>
<td>-0.033</td>
<td>-0.397</td>
<td>0.245</td>
<td>0.129</td>
<td>-0.100</td>
<td>0.414</td>
<td>0.352</td>
<td>0.302</td>
<td>0.156</td>
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<td>0.137</td>
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<td>-0.202</td>
<td>-0.361</td>
<td>0.482</td>
<td>-0.647</td>
<td>-0.223</td>
<td>-0.613</td>
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<td>-0.387</td>
<td>0.137</td>
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<td>0.642</td>
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<td>-0.100</td>
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<td>-0.650</td>
<td>-0.339</td>
<td>-0.357</td>
<td>-0.083</td>
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<tr>
<td>LMR-ML</td>
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<td>-0.072</td>
<td>-0.033</td>
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<td>-0.064</td>
<td>1</td>
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<td>-0.430</td>
<td>-0.083</td>
<td>-0.220</td>
<td>-0.188</td>
<td>-0.557</td>
<td>-0.186</td>
<td>-0.481</td>
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<td>0.642</td>
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<td>-0.583</td>
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<td>-0.272</td>
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<td>0.286</td>
<td>0.245</td>
<td>-0.202</td>
<td>-0.340</td>
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<td>0.528</td>
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<td>0.846</td>
<td>0.287</td>
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<td>0.227</td>
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<td>-0.358</td>
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<td>0.504</td>
<td>0.605</td>
<td>0.414</td>
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<td>-0.650</td>
<td>-0.188</td>
<td>-0.603</td>
<td>0.481</td>
<td>0.389</td>
<td>-0.492</td>
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<td>0.514</td>
<td>0.767</td>
<td>0.168</td>
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<td>BW-ML</td>
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<td>0.104</td>
<td>0.118</td>
<td>0.352</td>
<td>-0.223</td>
<td>-0.339</td>
<td>-0.557</td>
<td>-0.583</td>
<td>0.528</td>
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<td>0.180</td>
<td>0.514</td>
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<td>0.429</td>
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<td>-0.023</td>
<td>-0.083</td>
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<td>0.168</td>
<td>0.812</td>
<td>0.319</td>
<td>1</td>
</tr>
</tbody>
</table>

Bold: |correlation| >= 0.8, Italic: 0.8 > |correlation| >= 0.5
2.2.2.4 Variance analysis of the frequency type CoP parameters

The ANOVA addresses the investigation of the potential behavior of frequency-type CoP parameters between stance types. To meet the ANOVA criteria seven further participants had to be excluded from the study because some of their parameters appeared to be extremes compared to the rest of the cases especially in the ratio type parameters (Figure 2.5). The final number of samples used to the repeated measures ANOVA was 18. Multivariate test showed a significant difference between the three stance types (EO, EC, SL) (p<0.001) as expected. Mauchly’s test of sphericity was only violated in two parameters (ML directional largest amplitude and median frequency) that the robust ANOVA could still handle. On the basis of the within-subject univariate test, all parameters except MDF-AP are significantly influenced by stance types, as they significantly differ from each other in the three stance types. As regards pairwise comparisons (Table 2.6), in respect of some parameters (LA-AP, LMR-ML, MDF-ML, MPF-ML) bipedal stance with EO significantly differ from both bipedal stance with EC and SL stance, while the latter two stance types do not differ significantly from each other. In the most common cases (LAF-AP, LA-ML, MHR-AP, MHR-ML, SPR, BW-AP, MPF-AP) EO and EC stances do not differ significantly from each other while SL stance statistically significantly differs from both bipedal stances. In case of some other parameters (LAF-ML, BW-ML) all the three stance types significantly differ from each other.

![Box plots with extreme values and outliers (dots) in some of the ratio type parameters.](image)

**Figure 2.5.** Box plots with extreme values and outliers (dots) in some of the ratio type parameters.
Table 2.6. Pairwise comparisons in the frequency type parameters between stance types

<table>
<thead>
<tr>
<th>Measure</th>
<th>Mean Difference</th>
<th>Standard error</th>
<th>Level of significance</th>
<th>95% CI Lower Bound</th>
<th>95% CI Upper Bound</th>
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<td>LA AP</td>
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<tr>
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<td>1.814</td>
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<td>0.012</td>
<td>-0.293</td>
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<td>0.062</td>
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<td>-0.306</td>
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<tr>
<td>LA ML</td>
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<td>-7.783</td>
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<tr>
<td></td>
<td>EO SL</td>
<td>-0.163</td>
<td>0.068</td>
<td><strong>0.028</strong></td>
<td>-0.307</td>
</tr>
<tr>
<td></td>
<td>EC SL</td>
<td>0.023</td>
<td>0.093</td>
<td>0.807</td>
<td>-0.174</td>
</tr>
</tbody>
</table>

Bold: significant difference (p<0.05)
2.2.3 Discussion on CoP parameter reduction

The present research aims to reduce the number of the time-distance type and the frequency type CoP parameters by excluding those that contain redundant information, also replaced multiple parameters with summarizing parameters which based on the replaced parameters give a new and independent approach of the same phenomenon. Redundancy of the parameters were tested using correlation analysis. Independent parameters were selected and tested for sensitivity using variance analysis between different stance types. Based on the results the recommended CoP parameters are summarized in Table 2.7. The results and the recommendations are explained in this section.

Table 2.7. Recommended independent and sensitive CoP parameters

<table>
<thead>
<tr>
<th>Time-distance parameters</th>
<th>Confidence ellipse area (CE area)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Confidence ellipse axis ratio (CE axis ratio)</td>
</tr>
<tr>
<td></td>
<td>Path length</td>
</tr>
<tr>
<td></td>
<td>Maximum path velocity</td>
</tr>
<tr>
<td></td>
<td>AP-ML range ratio</td>
</tr>
<tr>
<td></td>
<td>Anterior (AP+) and Posterior (AP-) maximum deviations</td>
</tr>
<tr>
<td></td>
<td>Largest amplitude during balancing (LA)</td>
</tr>
<tr>
<td></td>
<td>Load distribution difference (LDD)</td>
</tr>
<tr>
<td>Frequency parameters</td>
<td>Frequency power ratios between low-medium and medium-high frequency bands (LMR, MHR)</td>
</tr>
<tr>
<td></td>
<td>Mean power frequency (MPF)</td>
</tr>
<tr>
<td></td>
<td>Spectral power ratio (SPR)</td>
</tr>
</tbody>
</table>

2.2.3.1 Time-distance type CoP parameters

Correlation analysis revealed significant correlations between many of the previously used variables in describing CoP motion, which makes them less suitable for variance analysis and introduces unnecessary redundancy in the CoP analysis based
evaluation processes. The standard deviations and CoP ranges show strong correlations (Table 2.2) in the respective directions, but this happens only when there are no large excursions in the CoP trajectory compared to the size of the confidence ellipse. This is due to the fact that the standard deviation (as well as confidence ellipse) describes the whole average of the measurement, while the ranges describe the extremes of the CoP trajectory. Usage of the ellipse axes together with the ellipse area contains redundant information. The present research showed that it would be a better approach to describe the area of the ellipse along with its kurtosis defined by the major-minor axis ratio.

*AP-ML range ratio* correlates with the ellipse axis ratio, but as before the range ratio describes the extremes, while axis ratio describes an overall statistic, thus both parameters may be used. The *AP ± maximum deviations* and corresponding angles tells the sizes of the largest excursions in the anterior and posterior directions. These reveal the largest CoP excursions relative to the average CoP, which might stay hidden using only the AP and ML ranges.

The comparisons of three trial types (bipedal stance with open eyes, bipedal stance with closed eyes and single leg stance) for the parameters used so far showed the same results that can be found in the literature [48]. Namely the values of the parameters rise but not significantly between EO and EC bipedal stance, whereas they rise significantly in single leg stance. *Path length* changed significantly between the two bipedal stance trials as well (Table 2.4). *AP maximum deviations* and 95% CE axis ratio presented the same behavior. *AP-ML range ratio* surprisingly not differs significantly between EO bipedal and single limb stance, but there is a significant deviation in the two trials compared to the EC bipedal stance. This is due to that the CoP overshoots in EO bipedal stance are equally small in both AP and ML directions while they are equally large during single leg stance, while in EO bipedal stance the sway and overshoots concentrated on the AP direction thus distorting the ratio.

### 2.2.3.2 Frequency type CoP parameters

It is worth discussing the result of correlation analysis along with the ANOVA results (Table 2.3 and Table 2.4). In most cases correlating parameters show similar behaviors to each other between stance types. Both *BW-AP* and *BW-ML* show a positive or negative correlation with many other parameters. *BW-AP* and most parameters
correlating with it significantly differ between bipedal stance types and single leg stance. Exceptions are LMR-AP and BW-ML. BW-ML also moderately correlates with four parameters and strongly with MPF-ML. However, in case of BW-ML the differences between stance types developed less systematically. CoP bandwidth contains common information with many other parameters, but does not precisely express postural characteristics, therefore BW parameters are recommended for exclusion.

MHR moderately correlate to themselves between AP and ML directions. This might be due to the rotational characteristic of CoP path presumed by Chiaramello et al.[49]. In EC condition, the CoP trajectory often becomes elongated as AP movements increase, thus the AP-ML range ratio increases [33]. The same behavior appears in the AP-ML spectral power ratio (SPR) of the CoP as oscillations are concentrated in the AP direction under EC condition (Figure 2.6). In SL condition the SPR becomes smaller than in the bipedal stances as the oscillations are more evenly distributed in AP and ML directions.

![Figure 2.6. Spectral power ratio between stance types](image)

The negative correlations between BW and the frequency band ratios (LMR and MHR) are reasonable as most of the frequency power shifted to the lower frequency bands the bandwidth narrows. A moderate negative correlation between MPF and LMR in AP direction has a similar explanation as MPF follows the high power frequencies.

MPF-AP and MDF-AP have such a strong correlation that only one of them should be sufficient to express the same phenomenon. It is interesting to note that in ML direction this correlation is not reflected. In the ANOVA results MDF-AP did not show any difference between stance types but MPF did in both directions. As MPF is also strongly correlated to BW in both directions and BW has already been excluded,
preserving MPF and excluding MDF is recommended in the reduction of parameter number.

The LA and LAF parameters were graphically analyzed as well. Scatter plots and convex hulls around the LA-LAF points for each stance type are presented in Figure 2.7. A and B. The size of the largest amplitude showed no correlation with the largest amplitude frequency (Table 2.5). As it can be seen on Figure 2.7/A the LAF-AP values scatter uniformly approximately between 0.1 and 0.8 Hz in AP direction irrespective to the size of the amplitude. In ML direction (Figure 2.7/B) the same applies to LAF in the approximate range from 0.1 to 0.9 Hz. In ML direction the LA parameter clearly shows a separation of SL stance from bipedal stances. Therefore LA is appropriate parameter to characterize different stance types and possibly impairment conditions as well. LAF does not tell trustworthy information on postural stability, therefore its usage is not recommended.

The preserved CoP parameters can be useful in clinical cases. LA shows the amplitude of the continuous sway which was unobservable using conventional distance type CoP parameters – such as ranges and summary measures – as 95% confidence ellipse sizes [33]. This largest amplitude describes more informatively the random
errors of one’s postural control than the summary measures and absolute ranges. \( \text{LMR} \) and \( \text{MHR} \) can numerically describe the deviations of frequency bands that were analyzed by Nagy et al. in [16]. These ratios may also reveal alterations in the frequency domain that was analyzed by time-frequency analysis before, e.g. on vestibular impaired patients [41]. \( \text{SPR} \) could be useful in cases when the full spectral powers are separately analyzed in AP and ML directions. Oliveira et al in [50] have reported changes on the spectral power distribution of pregnant women that could be described by \( \text{SPR} \). \( \text{MPF} \) is already a widely used CoP spectral parameter that is used in a variety of studies, e.g. to monitor the balance development of children [51], or to determine falling related balance problems of people with Parkinson’s disease [52].

The study collected many CoP parameters used in stabilometry and defined some new parameters as well. These parameters were subject to correlation analysis to select the independent ones of which sensitivity to indicate postural changes were also studied by the means of variance analysis. A list of sensitive and independent parameters (Table 2.7) are proposed with possible clinical application areas. The test-retest reliability of these parameters is yet to be studied.
2.3 Reliability of the selected CoP parameters

2.3.1 Methods of the reliability study

The present reliability study follows the Guidelines for Reporting Reliability and Agreement Studies [53] and the recommendations for reliability studies [54, 55] and general recommendations for stabilometry studies [12, 13].

The measurement procedure and the used equipment and ethical approval is identical to what is detailed in section 2.1.4. The participants repeated sessions that contained each stance types (EO, EC, SL) detailed. Each session was repeated ten times. Between sessions, there were 1- to 5-min resting periods according to the needs of the subject, where talking and sitting down was permitted to avoid the effects of monotony and fatigue. Subjects were asked to walk in the room between sessions, and they had to be standing for at least half a minute before the next session started to avoid dizziness caused by a possible drop in the blood pressure owing to the sudden change of stance to the standing position.

The sampling frequency of the force distribution plate was set to 100 Hz. The sampling duration was set to 60 seconds, as suggested by Carpenter et al. [31], from which the first 30 seconds was used in the 30-second long measurement intervals.

2.3.1.1 Participants

Initially, assuming some consistency, a minimum sample of 26 participants was required for observing reliability 0.6 (planning value) compared with 0.4 (minimum values) by one rater with ten trials per participant and with $\alpha = 0.05$ (significance level) and $\beta = 0.8$ (statistical power) [55].

Thirty healthy subjects were included in the study (22 males, 8 females; average age: $22.0 \pm 6.47$ years; average body height: $1.77 \pm 0.06$ m; average body mass: $73.0 \pm 14.76$ kg, BMI: $23.16 \pm 4.25$). The exclusion criteria included the following: a history of vertigo or dizziness, vestibular or neurologic disorders, uncorrected visual problems, sustained lower extremity injuries, spinal disorders, use of medications that influence the balance system, hearing loss, and acute/chronic ear infections.
2.3.1.2 Assessed parameters

The list of the assessed parameters in this study was the independent CoP parameter list that has been concluded in Table 2.7. The definition of each parameter can be found in Table 2.1.

2.3.1.3 Statistical analysis

The combinations of the 30 healthy participants and the three stance types (EO, EC, SL) are considered to be our test cases for reliability assessment, with ten repetitions in each case, which results in ninety test cases and 900 individual measurements. The reliability in each stance type for the 30- and 60-second measurement interval was evaluated by traditional indicators such as ICC, SEM and MDC. The 30-second interval was the first half of the 60-second measurement. Evaluation of the ICC was based on the recommendation of Fleiss [56], and evaluation of SEM is according to Laroche [19]. ICC(2,1) was chosen to assess the reliability using a two-way random effect model with absolute agreement for single measures [57].

ICC often results in wide confidence intervals for its results. The values of SEM and MDC are based on the ICC, and therefore, they cannot complement the ICC for reliability analysis sufficiently. Thus, the covariance of variation (CV) (the other term is the relative standard deviation) was also determined within subjects as the ratio of the standard deviation (σ) to the mean (μ), as in (1):

$$CV = \frac{\sigma}{\mu} \quad (2.1)$$

As Atkinson and Nevill [58] describes, CV is not suitable for all types of parameters, such as parameters whose value distributes around zero (e.g., LDD). They also clarify that the calculated mean CV does not reflect the repeated test error for all individuals, but only the ‘average individual’. Correspondingly, the CV compliance rate (CVCR) was considered to extend the ICC-based analyses and to measure the ratio of the test cases in which the CV is below a given value. Ruhe et al notes in their review that CV ≤ 0.33 is a commonly used cutoff value in the interpretation of ‘acceptable’ CV [13]. As a rule of thumb, this cutoff value for the CVCR was set to 30%. The newly introduced CVCR reliability measure was defined as follows:
CVCR = \frac{n_{cv<30\%}}{n} \quad (2.2)

where \(n_{cv<30\%}\) is the number of test cases in which CV \(\leq 30\%\), and \(n\) is the total number of test cases.


2.3.2 Results of the CoP reliability study

Thirty subjects performed ten consecutive standing trials using three stance types (EO, EC, SL). No measurements had to be excluded. The results of the reliability analysis can be found in Table 2.8 for the eyes open condition, Table 2.9 for closed eyes and Table 2.10 for the single leg stance. Each table demonstrates the reliability variables for each of the assessed CoP parameters in the 30-second and 60-second measurement intervals. The ICC values are usually consistent in the 30- and 60-second intervals. The path length consistently yields the best ICC values in all configurations (> 0.7 during each condition except for EO 30-second, where 0.67). The LDD consistently yields fair-to-good reliability, with ICC 0.49 - 0.54 in both bipedal stances. The CE area yielded fair-to-good reliability in most cases. The LA parameters yielded mostly fair-to-good reliability in the 60-second measurements, while poor when only the first 30 seconds were analyzed. The ICC of AP LA was fair to good in each of the stance conditions with the 60-second measurement interval. The MPF parameters showed fair-to-good reliability in the EC and SL measurements, except for AP MPF in the 30-second SL condition. Usually, the other frequency type parameters showed poor reliability based on the ICC. The maximum CoP velocity parameter behaved differently, as it showed better (fair to good) reliability during the 30-second intervals in both bipedal stances, while it showed poor reliability with the 60-second periods. The maximum velocity has poor reliability in both SL conditions.

The SEM% and CV values reflect similar results to the lowest values for the path length. CVCR negatively correlates with the CV, but in some cases (e.g., the path length) shows that while the mean CV is way below the arbitrary cutoff value of 0.3, still not every subject could meet this criterion of CVCR < 100% (Table 2.8).
Table 2.8. Reliability measures with eyes open bipedal stance for 30-second and 60-second measurement intervals

<table>
<thead>
<tr>
<th></th>
<th>EO 30-second</th>
<th></th>
<th></th>
<th></th>
<th>EO 60-second</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ICC</td>
<td>SEM</td>
<td>95% MDC</td>
<td>CV [%]</td>
<td>CVCR [%]</td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>CE axis ratio</td>
<td>0.36</td>
<td>0.94</td>
<td>2.6</td>
<td>34.22</td>
<td>53.33</td>
<td>0.22</td>
<td>0.9</td>
</tr>
<tr>
<td>CE area</td>
<td>0.39</td>
<td>67.16</td>
<td>186.15</td>
<td>53.65</td>
<td>6.67</td>
<td>0.43</td>
<td>65.44</td>
</tr>
<tr>
<td>Path length</td>
<td>0.67</td>
<td>61.81</td>
<td>171.32</td>
<td>15.86</td>
<td>93.33</td>
<td>0.75</td>
<td>100.74</td>
</tr>
<tr>
<td>Maximum velocity</td>
<td>0.43</td>
<td>32.17</td>
<td>89.16</td>
<td>35.1</td>
<td>40.0</td>
<td>0.13</td>
<td>192.93</td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td>0.2</td>
<td>0.67</td>
<td>1.87</td>
<td>38.93</td>
<td>30.0</td>
<td>0.23</td>
<td>0.64</td>
</tr>
<tr>
<td>LDD</td>
<td>0.49</td>
<td>7.41</td>
<td>20.54</td>
<td>61.66</td>
<td>6.67</td>
<td>0.54</td>
<td>7.21</td>
</tr>
<tr>
<td>AP LA</td>
<td>0.31</td>
<td>6.19</td>
<td>17.17</td>
<td>34.56</td>
<td>43.33</td>
<td>0.46</td>
<td>6.41</td>
</tr>
<tr>
<td>ML LA</td>
<td>0.33</td>
<td>5.59</td>
<td>15.5</td>
<td>36.44</td>
<td>43.33</td>
<td>0.37</td>
<td>5.9</td>
</tr>
<tr>
<td>AP+</td>
<td>0.21</td>
<td>7.14</td>
<td>19.8</td>
<td>37.04</td>
<td>43.33</td>
<td>0.23</td>
<td>8.86</td>
</tr>
<tr>
<td>AP-</td>
<td>0.35</td>
<td>5.05</td>
<td>13.99</td>
<td>31.87</td>
<td>50.0</td>
<td>0.15</td>
<td>14.82</td>
</tr>
<tr>
<td>AP MPF</td>
<td>0.27</td>
<td>0.08</td>
<td>0.21</td>
<td>38.61</td>
<td>26.67</td>
<td>0.33</td>
<td>0.06</td>
</tr>
<tr>
<td>ML MPF</td>
<td>0.22</td>
<td>0.11</td>
<td>0.3</td>
<td>41.46</td>
<td>16.67</td>
<td>0.3</td>
<td>0.09</td>
</tr>
<tr>
<td>SPR</td>
<td>0.13</td>
<td>4.23</td>
<td>11.72</td>
<td>83.34</td>
<td>0</td>
<td>0.21</td>
<td>4.63</td>
</tr>
<tr>
<td>AP LMR</td>
<td>0.22</td>
<td>7.55</td>
<td>20.93</td>
<td>74.82</td>
<td>3.33</td>
<td>0.23</td>
<td>11.73</td>
</tr>
<tr>
<td>AP MHR</td>
<td>0.2</td>
<td>6.53</td>
<td>18.11</td>
<td>51.22</td>
<td>0</td>
<td>0.36</td>
<td>5.61</td>
</tr>
<tr>
<td>ML LMR</td>
<td>0.15</td>
<td>8.5</td>
<td>23.57</td>
<td>87.44</td>
<td>0</td>
<td>0.15</td>
<td>12.66</td>
</tr>
<tr>
<td>ML MHR</td>
<td>0.39</td>
<td>4.97</td>
<td>13.78</td>
<td>44.54</td>
<td>3.33</td>
<td>0.42</td>
<td>4.5</td>
</tr>
</tbody>
</table>
Table 2.9. Reliability measures with the eyes closed bipedal stance for 30-second and 60-second measurement intervals

<table>
<thead>
<tr>
<th></th>
<th>EC 30-second</th>
<th></th>
<th></th>
<th>EC 60-second</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ICC</td>
<td>SEM</td>
<td>95% MDC [%]</td>
<td>CV CR [%]</td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>CE axis ratio</td>
<td>0.15 (0.07-0.3)</td>
<td>0.93 (42.9%)</td>
<td>2.57</td>
<td>32.51</td>
<td>53.33</td>
<td>0.17 (0.08-0.32)</td>
</tr>
<tr>
<td>CE area</td>
<td>0.42 (0.29-0.59)</td>
<td>66.99 (49.3%)</td>
<td>185.68</td>
<td>43.83</td>
<td>20.0</td>
<td>0.39 (0.26-0.56)</td>
</tr>
<tr>
<td>Path length</td>
<td>0.71 (0.6-0.82)</td>
<td>59.24 (15.8%)</td>
<td>164.2</td>
<td>14.14</td>
<td>100</td>
<td>0.74 (0.63-0.84)</td>
</tr>
<tr>
<td>Maximum velocity</td>
<td>0.48 (0.35-0.64)</td>
<td>24.07 (33.1%)</td>
<td>66.71</td>
<td>27.44</td>
<td>66.67</td>
<td>0.08 (0.01-0.2)</td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td>0.36 (0.23-0.52)</td>
<td>0.6 (34.6%)</td>
<td>1.66</td>
<td>32.44</td>
<td>46.67</td>
<td>0.42 (0.29-0.59)</td>
</tr>
<tr>
<td>LDD</td>
<td>0.51 (0.38-0.67)</td>
<td>6.15 (69.9%)</td>
<td>17.05</td>
<td>64.24</td>
<td>3.33</td>
<td>0.52 (0.39-0.68)</td>
</tr>
<tr>
<td>AP LA</td>
<td>0.36 (0.23-0.53)</td>
<td>6.69 (36.3%)</td>
<td>18.54</td>
<td>30.73</td>
<td>50.0</td>
<td>0.49 (0.35-0.65)</td>
</tr>
<tr>
<td>ML LA</td>
<td>0.4 (0.27-0.56)</td>
<td>5.42 (45.4%)</td>
<td>15.03</td>
<td>34.99</td>
<td>46.67</td>
<td>0.48 (0.34-0.64)</td>
</tr>
<tr>
<td>AP+</td>
<td>0.35 (0.23-0.52)</td>
<td>5 (32.4%)</td>
<td>13.86</td>
<td>29.23</td>
<td>63.33</td>
<td>0.29 (0.17-0.45)</td>
</tr>
<tr>
<td>AP-</td>
<td>0.34 (0.22-0.51)</td>
<td>5.74 (36.2%)</td>
<td>15.91</td>
<td>31.12</td>
<td>53.33</td>
<td>0.11 (0.04-0.24)</td>
</tr>
<tr>
<td>AP MPF</td>
<td>0.42 (0.29-0.58)</td>
<td>0.07 (32.8%)</td>
<td>0.21</td>
<td>32.58</td>
<td>43.33</td>
<td>0.43 (0.3-0.6)</td>
</tr>
<tr>
<td>ML MPF</td>
<td>0.43 (0.3-0.6)</td>
<td>0.11 (36.6%)</td>
<td>0.29</td>
<td>36.27</td>
<td>36.67</td>
<td>0.4 (0.27-0.57)</td>
</tr>
<tr>
<td>SPR</td>
<td>0.33 (0.21-0.5)</td>
<td>3.73 (80.1%)</td>
<td>10.33</td>
<td>70.48</td>
<td>0</td>
<td>0.36 (0.23-0.53)</td>
</tr>
<tr>
<td>AP LMR</td>
<td>0.08 (0.02-0.2)</td>
<td>6.07 (108.3%)</td>
<td>16.82</td>
<td>79.06</td>
<td>0</td>
<td>0.2 (0.11-0.35)</td>
</tr>
<tr>
<td>AP MHR</td>
<td>0.28 (0.17-0.45)</td>
<td>8.46 (67.7%)</td>
<td>23.45</td>
<td>51.85</td>
<td>0</td>
<td>0.36 (0.23-0.53)</td>
</tr>
<tr>
<td>ML LMR</td>
<td>0.18 (0.09-0.33)</td>
<td>4.56 (108%)</td>
<td>12.64</td>
<td>82.75</td>
<td>0</td>
<td>0.24 (0.13-0.4)</td>
</tr>
<tr>
<td>ML MHR</td>
<td>0.34 (0.22-0.51)</td>
<td>5.93 (56.1%)</td>
<td>16.43</td>
<td>49</td>
<td>3.33</td>
<td>0.46 (0.33-0.63)</td>
</tr>
</tbody>
</table>
Table 2.10. Reliability measures with an eyes closed bipedal stance for 30-second and 60-second measurement intervals

<table>
<thead>
<tr>
<th></th>
<th>SL 30-second</th>
<th>SL 60-second</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>CE axis ratio</td>
<td>0.19 (0.09-0.34)</td>
<td>0.92 (50.2%)</td>
</tr>
<tr>
<td>CE area</td>
<td>0.58 (0.45-0.72)</td>
<td>93.63 (34%)</td>
</tr>
<tr>
<td>Path length</td>
<td>0.73 (0.62-0.83)</td>
<td>167.8 (16.8%)</td>
</tr>
<tr>
<td>Maximum velocity</td>
<td>0.22 (0.12-0.38)</td>
<td>91.44 (46.6%)</td>
</tr>
<tr>
<td>AP-ML range ratio</td>
<td>0.24 (0.13-0.4)</td>
<td>0.6 (38.2%)</td>
</tr>
<tr>
<td>LDD</td>
<td>not applicable</td>
<td>not applicable</td>
</tr>
<tr>
<td>AP LA</td>
<td>0.29 (0.18-0.46)</td>
<td>8.76 (34.3%)</td>
</tr>
<tr>
<td>ML LA</td>
<td>0.3 (0.18-0.47)</td>
<td>5.13 (25.3%)</td>
</tr>
<tr>
<td>AP+</td>
<td>0.34 (0.22-0.51)</td>
<td>5.5 (23.6%)</td>
</tr>
<tr>
<td>AP-</td>
<td>0.23 (0.13-0.39)</td>
<td>8.45 (35%)</td>
</tr>
<tr>
<td>AP MPF</td>
<td>0.24 (0.13-0.39)</td>
<td>0.12 (35.4%)</td>
</tr>
<tr>
<td>ML MPF</td>
<td>0.46 (0.33-0.62)</td>
<td>0.14 (25.1%)</td>
</tr>
<tr>
<td>SPR</td>
<td>0.12 (0.05-0.26)</td>
<td>6.89 (231.1%)</td>
</tr>
<tr>
<td>AP LMR</td>
<td>0.2 (0.1-0.35)</td>
<td>3.6 (93.7%)</td>
</tr>
<tr>
<td>AP MHR</td>
<td>0.47 (0.34-0.64)</td>
<td>1.9 (44.8%)</td>
</tr>
<tr>
<td>ML LMR</td>
<td>0.12 (0.04-0.24)</td>
<td>1.38 (87.6%)</td>
</tr>
<tr>
<td>ML MHR</td>
<td>0.44 (0.31-0.6)</td>
<td>1.35 (42%)</td>
</tr>
</tbody>
</table>
2.3.3 Discussion on the CoP reliability study

The reliability of eleven CoP parameters was studied on 30 healthy young individuals. The novelty of the study is that not only bipedal stances with eyes open or closed conditions were tested in 30- and 60-second trials but also a single leg stance on the dominant leg was tested as well. A new reliability parameter, CVCR, was additionally used to assess the reliability. The results show that the path length yielded consistently the highest reliability, while the LDD and CE area yielded consistently fair-to-good reliability. A difference could be observed in the LA parameters, where the 60-second sampling interval provided fair-to-good reliability in every stance type, while poor reliability was achieved in 30-seconds measurements. A comparison between our results and other studies that addressed the reliability of common CoP parameters is presented in Table 2.11, which indicates similar results between studies.
Table 2.11. Comparison of reliability (ICC) of common CoP parameters with the literature

<table>
<thead>
<tr>
<th>Population (sample size)</th>
<th>Sampling</th>
<th>EO</th>
<th>EC</th>
<th>SL</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>CE area</td>
<td>Path length or mean velocity</td>
<td>Maximum velocity</td>
</tr>
<tr>
<td>Healthy young adults (10)</td>
<td>30 s</td>
<td>0.61 (0.08–0.89)</td>
<td>0.82 (0.57–0.92)</td>
<td>0.79 (0.45–0.94)</td>
</tr>
<tr>
<td>Healthy young adults (12)</td>
<td>60 s</td>
<td>0.43</td>
<td>0.53</td>
<td>0.46 / 0.53</td>
</tr>
<tr>
<td>Healthy adults (12)</td>
<td>20 s</td>
<td>0.78</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young adults (30)</td>
<td>3x30 s averaged</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Healthy young adults (16)</td>
<td>60 s</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Healthy young adults (44)</td>
<td>51.2 s</td>
<td>0.768 (0.647-)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Healthy adults (21)</td>
<td>60 s</td>
<td>0.859 (0.749-0.934)</td>
<td>0.710 (0.528-0.855)</td>
<td>0.960 (0.923-0.982)</td>
</tr>
<tr>
<td>Hip osteoarthritis patients (38)</td>
<td>54 s</td>
<td>0.52 (0.08-0.8)</td>
<td>0.85 (0.64-0.95)</td>
<td>0.76 (0.44-0.91)</td>
</tr>
<tr>
<td>Spinal cord injury patients (23)</td>
<td>51.2 s</td>
<td>0.64</td>
<td>0.89</td>
<td>0.94</td>
</tr>
<tr>
<td>Knee osteoarthritis patients (25)</td>
<td>3x10 s</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Study</td>
<td>Response Time</td>
<td>EC (0.00-0.60)</td>
<td>SL (0.09-0.99)</td>
<td>EC (0.3-0.6)</td>
</tr>
<tr>
<td>----------------------------------------------------------------------</td>
<td>---------------</td>
<td>----------------</td>
<td>----------------</td>
<td>--------------</td>
</tr>
<tr>
<td>Healthy elderly people (7)</td>
<td>30 s, 60 s, 120 s</td>
<td>0.22, 0.47</td>
<td>0.73, 0.77, 0.83</td>
<td>0.34, 0.09</td>
</tr>
<tr>
<td>People with musculoskeletal disorder (33)</td>
<td>30 s</td>
<td>0.33 (0.00-0.60)</td>
<td>0.84 (0.70-0.92)</td>
<td>0.64 (0.38-0.81)</td>
</tr>
<tr>
<td>Healthy (22) / Post-stroke (20)</td>
<td>10 s</td>
<td>0.63 (0.29-0.83), 0.35 (0.09-0.67) / 0.52 (0.13-0.77), 0.98 (0.95-0.99)</td>
<td>0.91 (0.8-0.96), 0.94 (0.98) / 0.82 (0.61-0.93)</td>
<td>0.91 (0.8-0.96), 0.94 (0.98) / 0.82 (0.61-0.93)</td>
</tr>
<tr>
<td>Current study</td>
<td>30 s, 60 s</td>
<td>0.39 (0.26-0.56), 0.43 (0.3-0.6)</td>
<td>0.67 (0.54-0.79), 0.75 (0.64-0.85)</td>
<td>0.43 (0.3-0.6), 0.13 (0.05-0.26)</td>
</tr>
</tbody>
</table>

EC and SL *maximum velocity* is not compared because it is not presented in any other studies. There were no studies to compare to for the EC and SL *maximum velocity.*
Few researchers studied the reliability improvement of CoP parameters by averaging several measurements [20, 21]. Although higher ICC values could be achieved by averaging, if the original standard deviation for the single measures is too large, the responsiveness for balancing alterations of the averaged parameter could degrade. Furthermore, the performance of repeated measurements is not always an option since time constraints might not allow it in clinical conditions, or it could be too exhausting for elderly people and people with impaired balance. Therefore, the averaging of the CoP parameters was not an option considered in this study.

The SEM and MDC values were consistent with the ICC values (Table 2.8, Table 2.9, Table 2.10). The main focus was on the relative SEM% values for relative evaluation and inter-parameter comparison since the parameters often have different dimensions. The smallest relative standard measurement errors were found for the *path length* in each stance type in conjunction with the ICC. The acceptance limit for SEM% (<15%) defined by Laroche et al [19] were not achieved for most of the parameters that were included in their study. This difference in the number of accepted parameters might be due to the deviations in the measurement procedure, because they downsampled the CoP data to 40 Hz before processing, while in the current study, 100 Hz was used. This difference could significantly influence the CoP parameters, mostly the frequency type parameters. Only the SL *path length* parameter achieved an SEM% that was lower than the limit defined in [19]. The explanation for this finding is that in the SL stance, the *path length* is significantly longer compared to the bipedal stances due to the increased postural sway. Because SEM% is the ratio of SEM and the mean, the otherwise similar standard deviations are relatively smaller compared to the increased mean, which results in smaller SEM% values.

The SL stance is an important test method in the analysis of the effect of the foot structure on the postural stability [35–37]. The SL stance could be truly useful in detecting better-than-average postural control, e.g., in research studies on sportsmen. On the other hand, the SL stance is often non-feasible for elderly and disabled people in clinical diagnostics.

Usually, the frequency type parameters yielded poor reliability. As described by [34], the frequency of the largest amplitudes in the CoP motions is inconsistent.
However, these large motions will result in the highest power frequency bins in the spectrum. If the frequency of these high-power frequency bins is inconsistent, then the resulting frequency-based CoP parameters yield large scattering in the values.

Most types of time-distance CoP measures can be evaluated in a simple manner: the larger the value is, the worst the balancing capability (CE area, path length, maximum velocity, anterior and posterior maximum deviation, LA). Although in some situations more stable persons might be able to afford more movement, instable persons control their movements more tightly. Ratio-type parameters are interpreted on a different scale and describe the orientation of the sway. These could be useful in clinical measurements when the monitored therapy affects the sway orientation or laterality. Otherwise, these are not recommended in cases where only simple and highly reliable assessments of the balancing quality are required. Anterior and posterior maximum deviations and AP LA and ML LA are extreme value parameters, just as the AP-ML range ratio is the ratio of the CoP path extremes. As such, they usually show poor reliability, but they are important to study because they can reflect the possibility of falling, which is an important field of study – especially in the elderly and in osteoarthritis patients.

Although LDD is not a CoP-based parameter, it is important, and it consistently showed fair-to-good reliability. Its usage can highlight the differences in the conditions of diminished limb health, especially in cases that involve the monitoring of unilaterally involved problems [14, 15].

The parameters that found to be reliable are also important from the biomechanical point of view. The path length accumulates all motion of the subject therefore indicating the average amount of motion that is a good measure of the balancing activity. The largest CoP deviation parameter is important while it can predict the limits of the comfortable and safe postural sway. This is important to measure in further studies on neurological or orthopedic alterations. On the other hand, the largest amplitude parameter describe the dynamics of the balance compensations, where transient behavior of the balancing control can be analyzed in the time-domain. Lastly the load distribution difference show the asymmetry of the load bearing capability of the lower limbs which is a very important orthopedic metric.
The present research might play a role in planning future stabilometry studies, because the reliability of several time-distance and frequency-based CoP parameters were studied on young, healthy individuals not only in bipedal stances but also in a single leg stance for 30- and 60-second measurement intervals. The limitations of the study are the lack of external validity to aged or diseased populations and the lack of analysis of factors such as sampling frequency or filter design.

2.4 Application of the determined parameter set

The reduced and validated CoP parameter set can be used for stabilometry measurements in several fields. During our studies we have used these reduced parameter list in multiple studies, including the study of the effect of orthopedic deviations on stabilometry and study the effect of virtual reality on stabilometry. The proposed parameter set was used to study the effect on postural balance of bad posture [60], knee osteoarthritis [61] and pes planus [62]. Detailed results can be found in the referenced papers. These parameters together proved to be useful in the above medical applications to study the postural balance and indicate deviations from healthy controls while revealing underlying reasons.

1. Contribution

The following parameters should be used to describe the motion of the center of pressure and deviations of the balancing capability for the purpose of stability tests:

- the path length of the foot center of pressure,
- largest anterior and posterior (forward and backward) deviation from the average center of pressure (Figure 2.3),
- the largest continuous amplitudes projected onto the axes (Figure 2.4),
- load distribution difference between lower limbs.

Related publications: [P1-6]
3 VALIDATION OF OPTICAL MOTION CAPTURE SYSTEMS

3.1 Introduction to motion capture system validation

Motion capture systems are widely used to measure human kinematics. These are multi-camera systems that track the 3D position of light reflecting or light emitting markers to track body motions based on the principles of stereophotogrammetry [63]. The use of low cost optical motion capture (OMC) multi-camera systems is spreading in the fields of biomechanics research [64] and rehabilitation [65]. OptiTrack is one of these relatively new names on the market [66]. There are custom made systems as well such as the one presented in Devecseri et al. [67]. These systems like high-end motion capture systems use retroreflective markers and triangulation to calculate marker coordinates. An important scientific criterion of their application is the identification of their accuracy, validity and reliability [68].

Extrinsic camera calibration is a common activity before OMC measurements. OMC software provide calibration quality characteristics whose acceptance is subjective (e.g. mean reprojection 3D error, residual mean error of the markers, and suggested maximum length for the reprojection rays). However, these are self-evaluated calculations of the OMC system. Therefore, researchers often conduct additional validation studies on the accuracy of motion capture systems.

3.1.1 State of the art

Different OMC systems are sometimes validated using Vicon camera systems (Vicon Motion Systems Ltd, Oxford, UK), which are regarded as the gold standard in scientific applications [69, 70]. Validation of OptiTrack camera systems compared to a Vicon system are presented in the literature [66, 71, 72]. In this case, marker coordinates and derived parameters are compared, measured by both the target and Vicon systems in simultaneous measurements. In each of these three studies, marker distances of marker rigid bodies were measured in different measurement volumes ([71]: 2 m × 2 m × 2 m, [66]: 2.5 m × 1.5 m × 1.5 m, [72]: different capture volumes with 10 m walkway). Carse et al. [72] established that there was an average 2.2% disagreement between the Vicon MX system (12 cameras, resolution not reported) and the OptiTrack
V100:R2 (8 cameras, resolution not reported) when measuring 65-140 mm marker distances (marker diameter: 16.5 mm)). Regarding reliability, the coefficient of variation for the marker distances was 0.3% for the Vicon MX (largest single error: 1.7 mm) and 2.3% for the OptiTrack system (largest single error: 18.23 mm). In the study referred to above, there were no ground truth measurements regarding marker distances, only the measurements of the OMC systems were compared. Hansen et al. [71] also validated an OptiTrack system (250e, 10 cameras, resolution: 832 × 832 pixel) comparing to a Vicon system (M2, 8 cameras, resolution: 1020 × 656 pixel) but also compared the marker distance measurements of both systems to the nominal distance of the markers (ranging from 64 – 500 mm, marker sizes 9 and 14 mm). In this research [71], root mean square errors of the measured distances were below 1 mm for both camera systems which fall within the reported average error range of the camera system calibrations (Vicon: 0.962 mm, OptiTrack 0.398 mm). Thewlis et al. [66] also compared a Vicon MX (12 cameras, resolution 1600 × 1280 pixel) system to an OptiTrack Flex V100R2 (12 cameras, resolution 640 × 480 pixel) in measuring a marker distance in which length is known (120 mm). They reported that although the OptiTrack system yielded higher errors compared to Vicon, none exceeded 1% neither in static nor in dynamic conditions. The authors also conducted another experiment where coordinates of markers on anatomical points were simultaneously measured and gait parameters were extracted for the comparisons of the systems [66]. The relative differences on gait parameters were mostly below 4°, only knee rotation showed 4.2° difference at 75% of stance phase. Based on their findings in comparing the accuracy of OptiTrack to Vicon systems, the above three papers concluded that the studied low-cost OptiTrack systems are suitably accurate for human motion analysis.

Another validation approach relies on measurements of the coordinate-based or relative displacement of a marker that is moved in a precisely known trajectory within small volumes, mostly moved by a micron resolution robotic device or a linear motion stage [73–75]. It is important that in these studies, accuracy was not compared to another system based on the same principles, but to another precise external reference. Aurand et al. [75] have studied the accuracy of a 42 camera OptiTrack Prime 41 system (4.1 megapixels) measuring the small relative displacement of a single marker (diameter: 15.9 mm). The measurement was performed in different locations of a large
measurement area \((10.4 \text{ m} \times 6.5 \text{ m} \times 2 \text{ m})\). 3D error was measured compared to a linear motion stage, which moved the marker with 5 µm off-axis error in a maximum 100 mm range. They concluded that the worst error of these relative motions is less than 200 µm in 97% of the 10.4 mm × 6.5 mm measurement area, and reaches 1 mm error only at the edges of the measurement area. It is important to note that although the measurements were performed in different locations of a large capture volume, only 100 mm of relative displacements were measured, as the origin was placed always in the starting position of the linear stage.

The research of Windolf et al. [74] on a Vicon 460 OMC system established that accuracy depends on multiple factors such as camera number and resolution, capture volume, and lighting conditions. These conditions influence accuracy even when the experiments are conducted in a 180 mm × 180 mm × 150 mm measurement area, and the trajectory of markers moved by precision robotic equipment studied. Eichelberger et al. [76] established similar conclusions on a Vicon Bonita system measuring marker distances in a 5.5 m × 1.2 m × 2 m capture volume. Based on the findings of the above studies [66, 71, 74, 76] it is evident that the accuracy of camera systems often requires scientific validation, even in single lab installations.

The validation methods described above can determine multiple aspects of accuracy and are well applicable to quantify static and dynamic accuracy, but are not suitable for measuring the absolute accuracy of larger capture volumes and the possible scale type errors of OMC systems. A scale type error results in coordinate deviation from the ground truth, linear to the distance from the origin of the OMC coordinate system, irrespective of where it is defined. While they are simple and suitable for dynamic validations, in case of techniques that measure relative distances of markers rigidly moved together [66, 69, 70, 72], the same scaling distortion applies to the coordinates of each marker in the marker cluster, therefore the scaling effect will not be revealed. Validation techniques using motorized motions can provide submillimeter resolution reference coordinates, but are only applicable in small volumes due to the construction of these robotic precision mechanisms [74, 75]. Even though Vicon systems are considered the gold standard in science, comparing the accuracy of other systems to it can be misleading: while they work based on the same principles of stereophotogrammetry, they have similar errors to low-cost systems [71], and can be
influenced by scale error as well. The previously studied errors in the literature include deviations in gait parameters compared to a reference system, distance error of relative marker displacements, error of relative marker distances or coordinate fluctuations of static markers, but there are no known methods reported that are capable of measuring absolute accuracy and quantify scale errors of OMC systems in capture volumes suitable for human motion capture using references established by a different approach than the validated system. It is hypothesised that by doing so a scale type error might be observed that was not addressed in previous validation studies.

3.1.2 Outline of the study

The aim of the present chapter is to introduce a novel method of validating the static absolute accuracy of OMC systems in larger measurement volumes and it characterizes scale errors of OMC systems exemplified on an 18 camera OptiTrack Flex13 (NaturalPoint, Corvallis, OR, USA) OMC system. Two approach are introduced using surveying measurements to validate the absolute accuracy of the OMC system in large measurement volume. The first approach uses an external planar reference network of marker locations defined by electro-optical distance measurements and repeatedly placed and measured marker coordinates compared to the measurements of the OMC system. The second approach uses micro-triangulation, an engineering-surveying method for ground truth measurement of the measured marker coordinates in the 3D space of the larger capture volume. A simple and easily replicable method to compensate scale errors of OMC systems is also introduced and its results are compared to the high-accuracy surveying method.

3.2 Methods

3.2.1 Camera system configuration

The experimental setup was employed in the Motion Analysis Laboratory of the Department of Mechatronics, Optics and Mechanical Engineering Informatics at the Budapest University of Technology and Economics in Hungary. The room where the OptiTrack system was installed has permanently darkened windows to avoid infrared
(IR) interference caused by sunlight. The evaluated system consists of 18 OptiTrack Flex13 cameras (NaturalPoint, Corvallis, OR, USA). The cameras were of 1.3 Megapixel resolution (1280 × 1024 pixel) and were equipped with stock lenses (5.5 mm focal length, 56° horizontal and 46° vertical field of view). Before the measurements, the focus of the cameras was adjusted where needed. The markers of the measurement setup were checked on the images of each camera and where the overall sharpness of the marker edges was visually out of focus, compensatory focusing was applied as described in the manufacturer’s supporting manual [77]. Other general manufacturer’s guidelines described in the supporting webpage [77] for optimal precision capture were followed in the camera placement and calibration process. The cameras were equidistantly placed in altering elevations around the room using wall-mounted consoles at 3 m above the floor (Figure 3.1). The field of view of each camera was aligned to ensure the largest possible overlapping in the capture volume for full body human motion capture in a 4 m × 2.5 m × 3 m area. Motive v1.10.3 software (NaturalPoint, Corvallis, OR, USA) was used for camera system calibration and capture.

### 3.2.2 Surveying control network for planar measurements

The present study has been conducted in cooperation with the Department of Geodesy and Surveying of Budapest University of Technology and Economics (Hungary). The independent control network used for 2D quality control of the OptiTrack system consisted of a 0.5 m raster grid with a dimension of 4 × 2.5 m, totalling 30 m³ of capture volume. In addition, 16 surveying reference points were fixed on the walls and camera consoles to support repeated measurements. These reference points enable the re-establishment of the control network at any time with millimetric accuracy. The grid points of the control network were set using a Leica TS15i 1” total station (Leica Geosystems AG, Heerbrugg, Switzerland) and aligned the local coordinate system to the grid orientation. The points were set based on infra-red electro-optical distance measurements using a mini prism with a sharp tip. First, the points were sat and then fiducial markers were stuck to the point locations (Figure 3.1). Although the mean error of the distance measurements using a prism and specified by the instrument manufacturer is 1 mm + 1.5 ppm [78], practice showed that a higher accuracy of approximately 0.5 mm can be expected when measuring short distances.
The control points were measured twice on different days and from independent stations, based on the previously established surveying reference points on the walls and consoles. The station coordinates were calculated using the instrument free station procedure applied to redundant observations. The root-mean-square error (RMSE) of the station coordinates remained below 0.5 mm for the two measurements. To further increase accuracy, the mini prism was operated using its automatic target recognition function, which ensures an angular accuracy of 1” for short distances. The final coordinates of the control points were determined by the mean coordinates between the two measurements.

Precision $P_g$ of the surveying measurement method is defined by

$$P = \sqrt{\frac{1}{nN} \sum_{i=1}^{N} \sum_{j=1}^{n} \left( (\bar{x}_i - x_{ij})^2 + (\bar{y}_i - y_{ij})^2 \right)},$$  

(3.1)

where $n$ is the number of measurement repetitions for each point, $N$ is the number of surveying points, $x$ and $y$ are the measured coordinates of the corresponding surveying points. For this study, we set $n = 2$ and $N = 54$. Furthermore, uncertainty $U_g$ of the surveying reference system is given by

$$U = k \cdot P = 2P \ (95.5\% \ CI),$$  

(3.2)

which measures the 95.5% confidence interval, $CI$, of the dispersion and equals two times the precision.

The complete setup with the camera arrangement and the surveying reference points is shown on Figure 3.2, where the values indicate the number of cameras pointing at the corresponding surveying points.
Figure 3.1. Establishment of the surveying reference network

Figure 3.2. Camera system layout and geodetic reference points (a), where the values indicate the number of cameras pointing at each location and real camera system setup (b)
3.2.3 Experimental procedure

3.2.3.1 Calibration of the OptiTrack motion capture system

The system was turned on 1 hour prior to the measurement and calibration to heat up as recommended by the manufacturer’s precision capture guide [77]. The camera system was calibrated using the standard CW-500 calibration wand manufactured by OptiTrack to be used for calibration; its nominal size is 500 mm. The true size of the calibration wand at 21 °C was also measured with an ATOS II Triple Scan MV320 (GOM, Braunschweig, Germany) high precision optical 3D scanner device for later comparison with our wand compensation results. The resolution of this optical scanner varies from 0.02 mm at 490 mm distance and 0.79 mm at 2000 mm distance from the scanner. The measured wand was placed at 500 mm distance from the scanner to fit in the viewing angle of the scanner and measured 499.75 mm wand size (0.05% deviation from nominal size).

The calibration wand was moved and rotated slowly in three dimensions of the measurement area, paying special attention in the target area where the markers were to be placed later. The wand was also moved in the edges of the volume and the camera field of views in order to help the software improve lens distortion. Calibration was refined until no untracked rays were present around the measured markers using default reconstruction settings, which are indicators of miscalibrated cameras. This condition was controlled during the measurements. An average calibration result reported a mean 3D error of 0.517 mm and suggested a maximum tracking ray length of 7.9 m. Based on our setup, the largest possible tracking rays are about 5.8 m long.

Figure 3.3. CW-500 calibration wand
3.2.3.2 Measurement with the planar surveying reference network

Measurements were taken using individual OMC system calibrations described above. Given that the OptiTrack system only detects IR light and the cameras are equipped with IR illumination, we used brand new OptiTrack retroreflective markers with a diameter of 8 mm for all the measurements. We placed the markers on the geodetic reference points with their bore pointing downwards with the highest possible precision to measure the reference positions by the OptiTrack system. However, this process includes considerable human inaccuracy on the placement of the markers. Therefore, four people performed the marker placement, and it was repeated 10 times per person to give a statistical basis for the process. Each person was asked to avoid a systematic order for the marker placement to prevent related errors. Therefore, we expected that the average among coordinates from the 40 repetitions for each calibrated reference point would reflect its true position. Precision $P_m$ for the marker placement over the 40 repetitions was defined by equation 3.1, considering $n = 40$ and $x$ and $y$ as the coordinates of the markers measured by the camera system. Likewise, uncertainty of the placement $U_m$ was given by equation 3.2 using the precision for the marker placement. The camera system calibration, repeated marker placements, and subsequent measurement cycles were conducted on separate days.

3.2.3.3 Measurement with direct micro-triangulation of the markers in 3D

Retroreflective markers were placed in the measurement volume in a $3 \times 3$ grid on the floor with about 1 m distances between the markers. Further markers at about 400 mm and 800 mm heights – at about the height of knee and pelvis – were placed above these markers (Figure 3.4) on the sitting surface and on top of the backrest of chairs. Knee and hip heights at 500 and 1000 mm have previously been used by Eichelberger et al. for both static (stationary markers) and dynamic (moving markers) validation [76]. The setup was completed in order to minimise marker occlusion by the chairs and to ensure sufficient coverage of the markers by at least six cameras. Three further markers were placed on the floor that were used to designate the reference frame of the OptiTrack system by assigning the origin, a point on axis $x$, and a point on plane $x$-$z$, labelling the points as ‘O’, ‘X’ and ‘Z’. The configuration was recorded for 30 seconds at 120 fps in the Motive software with calibration self-evaluation mean residual
error of markers: 0.517 mm and recommended residual for precision capture: 0.8 mm or below.

Marker coordinates were also measured by two Leica TS15i 1" total stations, which are high precision tools for engineering surveying measurements (angular accuracy: 1”). The measurement setup is shown in Figure 3.5. Sixteen surveying reference prisms were mounted on the walls and camera consoles to support reproducible measurements and to define an independent control network and reference frame for surveying measurements. The two total stations’ coordinates and orientation were determined for each measurement with free stationing using a built-in program based on the wall mounted control network. The average standard deviation of these station coordinates was 0.4 mm. After the total stations were oriented, whole circle bearings (WCBs) and zenith angles were measured directly with 1” accuracy. Two WCBs and two zenith angles were measured at each marker (Figure 3.6) because targeting the center of a sphere is uncertain. The mean WCBs and the mean zenith angles were used to compute the marker 3D coordinates using the micro-triangulation method also applied in the CERN (European Organization for Nuclear Research) by Vlachakis et al. [79]. All marker coordinates were determined from two individual measurements, thus statistics were possible to be derived from the results. Precision \((P)\) of the surveying marker 3D coordinates was defined by

\[
P = \sqrt{\frac{1}{nm} \sum_{j=1}^{m} \sum_{i=1}^{n} \left( (x_{ji} - \bar{x}_j)^2 + (y_{ji} - \bar{y}_j)^2 + (z_{ji} - \bar{z}_j)^2 \right)},
\]

where \(m\) is the number of markers (30), and \(n\) is the number of independent measurements (2). Uncertainty \((U)\) of the surveying reference measurements on a 95% confidence interval is defined by \(U=2P\)\text{[95\% CI].}
**Figure 3.4.** Marker locations in the measurement volume relative to the cameras

**Figure 3.5.** Measurement setup with illustrations of surveying measurement definitions
3.2.4 Analysis of absolute accuracy and wand compensation

The absolute accuracy of the system was characterized by the RMSE of distance between the surveying reference coordinates of the markers and the coordinates measured by the OptiTrack camera system. In the planar measurements mean of the repeatedly placed and measured coordinates were used. As both measurements are performed on the once placed markers in the 3D measurements, human error of marker placement into a specific location does not apply. RMSE was calculated as

$$RMSE = \sqrt{\frac{1}{m} \sum_{i=1}^{m} \left( (x_{i}^{geo} - x_{i}^{opt})^2 + (y_{i}^{geo} - y_{i}^{opt})^2 + (z_{i}^{geo} - z_{i}^{opt})^2 \right)}$$

(3.4)

where geo refers to the coordinates of the reference points determined by the surveying method, opt refers to the coordinates by the OptiTrack system represented in the common coordinate system, and \( m \) is the number of measured points. Coordinate \( z \) is zero in the planar measurements.

This comparison requires the surveying and the OptiTrack coordinates to be represented in the same coordinate system. The marker coordinates from both systems were translated by the corresponding coordinates of a selected marker aligning the
origin of the two systems (Figure 3.5). Angular errors of the OptiTrack coordinate system were compensated using homogeneous coordinate transformations on the marker coordinates in the order of the rotations around axes X, Y and Z. The angle of each rotation was established by minimizing the RMSE error. After the rotations the coordinate systems were aligned.

The residual error was decreased using a third independent surveying measurement by scaling the OptiTrack coordinates in order to minimize the following expression:

\[
\sqrt{\frac{1}{m} \sum_{i=1}^{m} \left( (x_i^\text{geo} - x_i^\text{opt} \times \text{scale})^2 + (y_i^\text{geo} - y_i^\text{opt} \times \text{scale})^2 + (z_i^\text{geo} - z_i^\text{opt} \times \text{scale})^2 \right)}
\]

(3.5)

where \text{scale} was decreased from 1 in steps of 0.0001 until the minimum result of the expression was reached. The use of a third, independent surveying measurement for compensation was important to have the scaling compensation independent of the original reference measurements that will be used to compare other scaling methods as well. The earned scaling factor was equal to the suggested scaling factor of the size of the particular calibration wand applied for precise large volume capture. The earned scaling factor was set in the measurement software as custom wand size and the calibration recording was reprocessed and the take with the markers was reconstructed with the compensated calibration. The new coordinates were compared again to the original surveying reference coordinates.

In addition, a correlation analysis was performed before and after compensation between the distance from the origin and the measured deviation to justify the scaling nature of the observed errors and their compensation.

3.2.5 Tape compensation

The above described method was meant to precisely identify the necessary scaling factor and to show the homogeneous effect of scale errors on the coordinates, although it is difficult to replicate without proper tools. Therefore, a simpler and less precise, easily reproducible test was also conducted using a surveying tape measure and two single markers glued onto the floor on the distal ends of the measurement area (but
not on the very edge to keep sufficient camera coverage of the markers, 4.72 m). Measuring larger distances with a less accurate tool to identify scaling errors reduces the effect of inaccurate measurement, as the scaling error increases on larger distances. The distance of the two markers was repeatedly measured with the tape measure (Figure 3.7) and averaged, while also measured by the motion capture software after a previously described precise calibration. This method is also based on surveying principles but requires no special instrumentation.

Figure 3.7. Measurement with geodetic tape measure

3.3 Results

3.3.1 Accuracy of the surveying reference system

3.3.1.1 Planar reference network

The precision (P in equation (3.1)) of 0.379 mm was obtained from the repeated geodetic measurements on the planar surveying control network (Figure 3.1). The
largest deviation of the geodetic measurement from the mean geodetic coordinate was 0.836 mm for a single measurement. This precision results in an uncertainty $U_x = 0.758$ mm of the reference point network. Therefore, smaller deviations of the measured marker coordinates from the reference points are not credible in the 95.5% confidence interval.

The marker placement precision of the individual marker placements and measurements is $P_m = 0.719$ mm, with the corresponding uncertainty $U_m = 1.438$ mm for each marker in the 95.5% confidence interval. This error is due to the repeated manual placement of the markers. However, this uncertainty does not influence our system evaluation, because we did not consider a single measurement but the average of the 40 measurements.

### 3.3.1.2 Accuracy of the surveying reference system in the 3D measurements

In this setup the markers are only placed once and measured twice with independent surveying measurements. Precision ($P$ in equation 3.3) of the independent surveying measurements was 0.37 mm while the maximum 3D deviation from the averaged coordinates was 0.67 mm for one marker. Therefore, uncertainty ($U$) of the surveying reference measurements is 0.75 mm on a 95% confidence interval.

### 3.3.1.3 Accuracy of the tape measurement

The standard deviation of the six tape measurements was 0.41 mm and the averaged distance was 4721.17 mm.

### 3.3.2 Absolute accuracy of the OptiTrack motion capture system setup

#### 3.3.2.1 Planar measurements

Absolute accuracy describes the difference between the reference and measured coordinates of the markers obtained from surveying and OptiTrack coordinate systems, respectively. Our initial results after coordinate system alignment showed significantly larger deviations than the uncertainty of the reference measurements. The planar deviations were characterized by an RMSE of 1.735 mm and largest deviation of 3.072 mm.
3.3.2.2 3D measurements

The 3D deviations were characterized by an RMSE of 1.82 mm and a maximum 3D deviation of 3.34 mm. Deviations between the marker and reference coordinates showed a strong correlation with the marker distance from the reference frame origin, with a correlation value of 0.81 in the planar measurements and 0.8552 in the 3D measurements.

Error optimization (Figure 3.8) yielded a scale value of 0.9992 in both approach with an improved RMSE of 0.967 mm in the 2D approach and 0.77 mm in the 3D approach. The correlation between the distance from origin and the errors 3D decreased to 0.36 in the planar approach and 0.209 in the 3D approach. The yielded RMSE is mainly covered by the also submillimetric uncertainty \( (U) \) of the surveying measurements. Error vectors are illustrated in Figure 3.9 and Figure 3.10 in the original and in the downscaled conditions, in a hundredfold magnification for better visibility.

3.3.2.3 Tape measurements

The simply reproducible and less precise measurement with tape measure earned similar downsampling results of 0.99958 as the distance of the two markers measured by the OptiTrack system was 4723.15 mm (averaged for the 30-second long measurement) and the averaged result of six tape measurements was 4721.17 (SD: 0.41) mm.

![Figure 3.8. RMS error and error correlation with distance from origin in the iterative optimization of the scale factor](image-url)
Figure 3.9. Planar deviation vectors between the average detected marker coordinates and the geodetic references. Results before the wand calibration optimization (a) and after applying the optimal scale factor (b). Error vector magnification was set to 100 for better visibility.

Figure 3.10. 3D error vector in the original (a) and the optimally downscaled (b) condition. Error vector magnification was set to 100 for better visibility.

3.3.3 Wand compensation

Based on the determined scaling factors by three different type measurements, wand size can also be compensated; this should be applied in the motion capture software for more precise measurements. The different measurements resulted in slightly different scaling factors and compensated wand sizes (Table 3.1). Each method
suggests scaling down the virtual space in order to better match surveying coordinates and to reduce distance dependent deviations, as shown in the correlations in Table 3.1. The fact that the surveying method earned higher precision compared to the high precision 3D scanner can be attributed to the fact that the RMSE with each wand size was compared between the surveying coordinates and the coordinates measured by the OMC system. The surveying coordinates can be influenced by errors of the 0.75 mm uncertainty of the measurement, and the surveying scale compensation was optimised for this precision.

**Table 3.1. Scaling factors and corresponding compensated wand sizes**

<table>
<thead>
<tr>
<th>Measurement / compensation technique</th>
<th>Scaling factor</th>
<th>Compensated wand size (mm)</th>
<th>Wand size difference (mm)</th>
<th>Compensated RMSE (mm)</th>
<th>Error correlation with distance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Original results (2D)</td>
<td>1.0</td>
<td>500</td>
<td>0</td>
<td>1.735</td>
<td>0.81</td>
</tr>
<tr>
<td>Original results (3D)</td>
<td>1.0</td>
<td>500</td>
<td>0</td>
<td>1.82</td>
<td>0.855</td>
</tr>
<tr>
<td>Optical scanner</td>
<td>0.9995</td>
<td>499.75</td>
<td>0.25</td>
<td>1.00</td>
<td>0.485</td>
</tr>
<tr>
<td>Surveying (2D)</td>
<td>0.9992</td>
<td>499.60</td>
<td>0.40</td>
<td>0.967</td>
<td>0.36</td>
</tr>
<tr>
<td>Surveying (3D)</td>
<td>0.9992</td>
<td>499.60</td>
<td>0.40</td>
<td>0.76</td>
<td>0.209</td>
</tr>
<tr>
<td>Tape measure</td>
<td>0.99958</td>
<td>499.79</td>
<td>0.21</td>
<td>1.14</td>
<td>0.608</td>
</tr>
</tbody>
</table>

**3.4 Discussion**

Large volume absolute accuracy and scaling errors of OMC systems have not been studied earlier. The aim of the present research is to analyse the scaling error and absolute accuracy of OMC systems by means of engineering surveying methods. Methods to compensate for the observable absolute inaccuracy are presented. Two method represents high accuracy using sophisticated instrumentation in planar and 3D measurements, while another simpler method can be performed easily with a tape measure. Observed errors are attributed to the scaling error of the calibration of the OMC system. The RMSE of the studied OMC system compared to surveying coordinates was managed to be reduced while its correlation with the distance also dropped, which justifies the elimination of the scaling error (Table 3.1). High precision 3D scanner measurement (0.02 mm resolution) and simple tape measurement of large
distances (4.72 m) in the capture volume resulted in the assessment of similar scaling errors and the possibility of compensation (Table 3.1).

The studied scaling error does not affect the well-known dynamic errors caused by e.g. the changes of camera coverage of the marker and its effect on the 3D reconstruction of the coordinates by reprojection rays during dynamic motion. We presented a static measurement and the studied scaling error causes a spatial distortion in the measurement area with coordinate errors linear to their distance from the origin of the reference frame. Other static and dynamic errors demonstrated in the literature [66, 71, 72] superimpose onto the scaling errors. Therefore, this spatial distortion applies to the dynamically moving markers, but as a predictable distortion and not as a random coordinate noise. Without compensation of a similar error as the presented (0.08%), for instance the nondeterministic 1.7 mm dynamic error of a Vicon MX system presented by Carse et al. [72], could be added to a 2.5 mm scaling error at a 5 m distance from the origin. The experiment demonstrates that the scaling error applies similarly in every 3D direction by pointing the error vectors towards or away from the designated origin of the coordinate system (Figure 3.9 and Figure 3.10) depending on upscaling or downscaling errors. Although camera number influences the precision of OMC systems [74, 75], the observed scaling error is not directly influenced by the number of cameras when sufficient camera coverage is provided. Consequently, the effect of scaling error is well distinguishable from other error sources.

Compared to relative distance measurements [66, 69, 70, 72], the proposed method is static and can be more complicated and expensive using the surveying total stations or less precise using the tape measure. On the other hand, the proposed method can characterize and compensate scale errors while the relative distance measurements cannot, even though these errors are magnitudes higher than the submillimeter errors that relative distance measurements usually detect. The micron accuracy of the reference system in the computer-controlled motorized relative displacement measurements [74, 75] might be superior compared to surveying, but these systems can move in a very small measurement volume of few hundred millimetres (Aurand et al. [75]: 100 mm linear, Windolf et al. [74]: 180 mm × 180 mm × 150 mm) and the observed scaling error is undetectably tiny in these small measurement volumes. Although they are properly suitable for accuracy validations of these small volumes,
they cannot be used for measurement volumes with distances of several meters, e.g. 4 m × 2.5 m × 3 m of our measurement volume. Aurand et al. [75] did perform their measurements in a large capture volume (10.4 m × 6.5 m × 2 m) in different places but the actual measured distance from the origin was always only 100 mm. Comparisons of the accuracy of an OMC system to a high-end OMC system should only be made when the errors of the high end system are also known and its possible scale error is already compensated. Consequently, all of the different validation methods should be applied together to characterize errors of OMC systems in as many aspects as they complement each other.

The ultimate purpose of the presented scale compensation is to indirectly yield the precise size of the calibration wand and to adjust the wand size in the measurement software of the OMC system for calibration. Directly measuring the distance of the markers at a theoretical 500 mm distance with the required accuracy might be challenging and requires a special measuring device as standard callipers have 150 mm measuring range and 0.02 to 0.05 mm accuracy. A 0.05 mm error on a 100 mm calibration wand could produce a 2.5 mm absolute error at a distance of 5 meters. When a level of precision is required for the measurements the useful capture volume with sufficient accuracy can be determined and enlarged by the proposed calibration, which is especially true in large capture volumes with possible scaling errors. OMC manufacturers (such as the manufacturer of the presently validated system, NaturalPoint) provide factory-calibrated ‘micron series’ calibration wands for precise measurements, but they are recommended for small measurement volumes. Their factory-calibrated custom size can be easily invalidated by just touching any of their markers [77]. The presented method with the tape measuring large marker distances (e.g. the currently applied 4.72 m) can be an easy indicator if the circumstantial factory calibration of these small wands is invalidated.

The effect of a scale type error of similar magnitude than the above would not affect significantly the results of actual biomechanical analyses such as gait analyses. The common errors in biomechanical motion analysis such as marker misplacement [80] or soft tissue artefact [81] are usually of similar or higher magnitudes, but the scale error affects each marker on the subject almost equally. This way their relative positions are less affected. It is like slightly moving the whole scene together when the subject is
away from the origin of the reference frame. The scale error might have a more noticeable effect when the OMC system is used as position feedback for a robot with high spatial position resolution.

The OMC system validation methods in the present research can be used to characterize the absolute accuracy of large capture volumes, which was an unsolved and ignored problem of previous validation studies and received little attention. The proposed methods are based on different principles compared to the validated systems therefore unaffected by the studied scale type error. The engineering-surveying technique characterized the 3D nature of these errors and the results could be used to compensate for the calibration wand size of individual configurations. The simpler and less precise method using tape measure earned similar results (Table 3.1) in scaling compensation compared to engineering surveying and to a direct wand size measurement with a high precision 3D scanner. This method can be an easy indicator of inaccurate calibration or invalidation of a high precision calibration wand. These absolute accuracy measurements can complete the previous OMC validation techniques for characterizing an error type that has not previously been studied. High precision of an OMC system is a key in the validation of other motion capture measurement systems using the OMC system as ground truth measurements.

2. Contribution

The static error of stereo-photogrammetry based motion capture systems can be determined using engineering surveying methods, whereby scaling error can be defined by simultaneous measurements of spatial position (different heights) of widely spaced markers in a common base coordinate system with the motion capture and a surveying systems, thus the system can be calibrated.

Related publications: [P7-8]
4 AUGMENTED REALITY BASED MOTION CAPTURE

This chapter introduces the idea and validation of an alternative motion capture system based on Augmented Reality markers that can be a cheap alternative compared to the expensive multi camera motion capture systems.

4.1 Introduction to Augmented Reality based motion capture

Gait analysis is the instrumented systematic study of human motion for measuring body kinematics and dynamics, and is used in medicine and biomechanical research to assess and treat individuals with impaired walking capabilities [82] or to improve sports performance [83]. A typical optical based motion capture gait laboratory has several cameras placed around a walkway. The subject has markers located at anatomical landmarks (Figure 4.1 a) or rigid groups of markers are applied to the body segments (Figure 4.1 b) [63]. In the latter case, the anatomical points are calibrated as virtual markers in the coordinate systems of the rigid marker clusters [63]. Trajectories of the markers or the position and orientation of the rigid bodies are calculated from the several camera pictures by the system using stereophotogrammetry [63]. The motions of the underlying bones are estimated to yield the joint kinematics. These motion capture camera systems are expensive, therefore there is a constant demand for more affordable gait analysis solutions with similar accuracy.

Figure 4.1. Gait analysis with markers on anatomical landmarks (a) and rigid marker clusters on body segments (b).
One of the current trends in gait analysis is the use of low-cost motion sensors based on inertial measurement units (IMUs) which combine sensor data from accelerometers and gyroscopes. These sensors are attached to body segments and measure the orientation of the segments. High precision orientation estimation of the IMU modules are possible due to advanced sensor fusion and filtering. Often a constrained biomechanical model is used to estimate body kinematics from sensor orientation data [84]. Using properly tuned constrained models and precise orientation tracking of IMU sensors gait analysis or other predefined motion types can be reliably measured [85, 86]. Whereas, if the constrained model is not accurate the measurement results can be biased, e.g. a commercial IMU based motion analysis system proved to be reliable on adults [86], but shows significant bias on children as calculates the model with adult leg lengths [87]. While these systems are affordable and mobile, they have limitations. Direct position tracking of the sensors is only possible by continuous or periodic integral drift corrections or zero speed update [88, 89] as accelerometer sensor readings contain noise which is exponentially accumulated in the integrated position data. An example for zero speed update is to zero out the estimated velocity when the foot is predicted to be on the ground during a gait trial [88]. To overcome the integral error, another common solution is the regression of the position data to zero, thus eliminating the error due to integrated errors of sensor drift [88]. Consequently, inertial systems work well on periodic motions but are less suitable for the absolute position tracking of objects, and the joint kinematics of a motion analysis highly depends on the constrained biomechanical model.

There are initiatives where open source solutions are provided to replicate the stereophotogrammetry based functionality of motion capture systems with consumer grade cameras. Jackson et al. [90] offers a complex solution for necessary camera calibration and the synchronization of video inputs from multiple cameras. This approach is based on stereophotogrammetry, where the identifiable points of the tracked object have to be seen from different angles by multiple cameras. Another image processing approach is homography, which relates the transformation between two planes [91]. This is used in photography for panorama picture stitching or perspective correction and is also used in augmented reality (AR) to estimate camera pose from coplanar points and vice versa. It can identify rotations and translations (3D kinematics)
of an AR marker relative to the camera focus point and the image plane by how the corners of the known geometry marker appear on the recorded image. Compared to continuously drifted or zero corrected IMU-s, the 6 degree of freedom tracking of AR markers make them possible to track the absolute position of external objects [92] and body segments if attached to them. Compared to stereophotogrammetry based alternatives [90], AR marker based tracking can work with one camera, although in this case the movement direction can be limited (e.g. treadmill walking).

AR was mostly mentioned so far in motion studies as a part of therapies [93], but not for the purpose of biomechanical motion tracking. Ortega-Palacios et al. describe a gait analysis system with augmented reality, but the localization of infra-red LED (light emitting diode) markers is still processed by stereophotogrammetry [94]. Sementille et al. used actual augmented reality markers to track the position of joints on a very simplified anatomical model [95]. None of the above research works validated the data acquired using a conventional motion analysis system.

The first aim of the present research is to present a novel approach for gait analysis with a single commercial action camera using augmented reality markers based on the approach of tracking body segments by marker rigid bodies [63]. Therefore, no simplification of the anatomical model is required, a full six degree of freedom kinematic analysis of each body segment and joint is possible using conventional or open-source motion analysis solutions such as OpenSim (NIH Center for Biomedical Computation, Stanford University, http://opensim.stanford.edu/).

The second aim of the present research is to validate a possible implementation of the proposed approach by simultaneous measurements with a conventional motion capture system on treadmill gait trials of healthy subjects of varying age at different walking speeds, followed by comparing the coordinates of the tracked virtual anatomical points and calculations for comparing angular and spatial gait parameters.

### 4.2 Methods

This section firstly describes the technical details of the proposed system. Secondly, the validation method of the system is described, which compares the accuracy of the AR marker system to a conventional optical motion capture system.
4.2.1 Description of the proposed Augmented Reality based motion capture system

4.2.1.1 Experimental procedure with the proposed system

The measurement protocol with the exemplifying implemented AR marker system has been registered and openly accessible in an online protocol description [96] with further illustrations. Gait trial with the present system starts by fixing specified AR markers onto the corresponding body segments of the subject using wide elastic bands to minimize soft tissue artifact [97]. All the markers have to be visible from the same direction during the complete trial, from where the camera is set up. The camera was set up about 1.5 meters behind the subject in order that each marker is visible on the camera in the whole movement range of the subject (Figure 4.2). In the present experiment the coordinate system was camera centered, so only the direct inaccuracies of the markers can be measured. The coordinate system could be arbitrary using another AR marker seen by the camera which defines the coordinate system position and orientation. In this solution, the exact orientation and position of the camera is irrelevant as long as each marker is well visible, thus the camera could also be a handheld smartphone. The drawback of this approach would be that the position and orientation detection error of the markers relative to the reference marker becomes multiplied compared to camera centered solution.

![Measurement setup of the AR marker based gait analysis.](image)

**Figure 4.2.** Measurement setup of the AR marker based gait analysis.
Before the measurement, anatomical landmark calibration has to be performed by palpation and with the help of a calibration pointer equipped with another AR marker (Figure 4.3). This procedure “teaches” the system the location of the indirectly tracked anatomical landmarks relative to their corresponding - directly tracked - AR markers using homogeneous coordinate transformation. Anatomical calibration is recorded by the camera and care must be taken so that the marker of the pointing wand and the calibrated body segment are well visible by the camera. Each anatomical point specified by the marker set has to be pointed with the pointing wand on the video. The calibration process takes about 1 to 2 minutes. Calibration is followed by gait trial on a treadmill for the desired time. The calibration and gait trial video files are processed offline by the image processing software where the frames of pointing to anatomical landmarks are selected manually. After this manual post-processing, a file with the calculated marker trajectories during the trial is available in a standard .trc file format. In the present experiment, the file is opened by a custom Matlab script which can perform calculations on marker trajectories and invokes a third party open-source biomechanical analyzing software (OpenSim) to calculate angular gait parameters.

The above procedure details the tested proof of concept implementation that we used. Further work should be invested in the optimization of the procedure where anatomical calibration and measurement evaluation is real-time (no post-processing), and the whole measurement could be performed even on a smartphone with a high-resolution camera and sufficient processing power.
Figure 4.3. Calibration of anatomical points using the calibration pointer.

4.2.1.2 Acquisition system

The accuracy of AR marker pose estimation depends mainly on the quality of camera calibration, which eliminates optical distortions and sets the resolution of input images and marker size on the image in pixels. Therefore camera calibration is an important technical aspect of image processing and system accuracy, but does not form part of the conducted measurements. Camera calibration needs to be done only once when configuring system parameters. During the measurements there is no need to deal with camera calibrations. For the AR marker detection algorithm, a high shutter speed is important to avoid unrecognizable blurry images at faster motions. From the viewpoint of gait analysis, the highest possible frame rate (fps) is required for high temporal resolution. It is also essential for the camera to have a fixed zoom and fixed focal length (disabled autofocus) because camera calibration is valid for fixed values of these parameters.

Several cameras have been tested and calibrated with different settings (Table 4.1), but only the results with the setup that proved to be the optimal in terms of the
above requirements are described in the thesis, which is a GoPro Hero5 Black action camera (GoPro, Inc, San Mateo, California, USA) set to 2.7k resolution at 50 fps with 1/200 shutter speed in linear mode. The linear mode of this camera runs a factory calibrated image undistortion on the device and the recording will be free from optical distortions; only focal length is the information required from camera calibration performed in this mode.

**Table 4.1.** Tested cameras and calibrated camera parameters

<table>
<thead>
<tr>
<th>Camera</th>
<th>Resolution*</th>
<th>Frame rate (fps)</th>
<th>Shutter speed</th>
<th>Focal length (in pixels)</th>
<th>Distortion parameters**</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kinect v2 (color video recording)</td>
<td>FullHD (1920x1080 pixel)</td>
<td>30</td>
<td>cannot be set</td>
<td>1034.68</td>
<td>k1: 0.0312, k2: -0.0450, k3: 0.0049</td>
</tr>
<tr>
<td>GoPro Hero 4 Silver</td>
<td>FullHD (1920x1080 pixel), narrow mode</td>
<td>60</td>
<td>cannot be set</td>
<td>1641.94</td>
<td>k1: -0.2971, k2: 0.1752, k3: -0.0755</td>
</tr>
<tr>
<td>GoPro Hero 5 Black</td>
<td>2.7k (2716x1524), linear mode</td>
<td>50</td>
<td>1/200</td>
<td>1483.71</td>
<td>k1: 0, k2: 0, k3: 0</td>
</tr>
<tr>
<td>GoPro Hero 5 Black</td>
<td>4k (3840x2160), wide mode</td>
<td>25</td>
<td>1/100</td>
<td>1775.89</td>
<td>k1: -0.2534, k2: 0.0894, k3: -0.0167</td>
</tr>
</tbody>
</table>

*Modes in GoPro cameras refer to the field of view option of the device; Kinect v2 has only a fixed wide field of view

**p_1 and p_2 distortion parameters are equal to 0 in each setup. Description of k and p distortion parameters are described in [98].

**4.2.1.3 Image processing**

Camera calibration was performed by OpenCV using a chessboard pattern [98]. The processed video frames are undistorted with the OpenCV undistort function before the tag detection algorithm is called (it has no effect when the linear video mode is used with the GoPro).

AR marker detection and identification are performed by the Apriltag algorithm using the 36h7 marker tag family [99]. Position of the detected tags and the rotation matrix with respect to the camera are given by the Apriltag algorithm using homography transformation which is available online.
(https://april.eecs.umich.edu/software/apriltag.html). The theory as well as the validation of the homography based orientation and pose estimation of the Apriltag algorithm can be found in the original paper of Olson [99]. In their validation on generated ground truth images, the angular error of the markers is less than 0.5° until about 75° off axis angle. The achievable accuracy of marker tracking in their results is significantly higher than with the more widespread ARtoolkit framework [99, 100], which they graphically present [99]. Another experimental validation of Apriltag’s marker tracking accuracy has been conducted by Pfleging et al. with a motion capture system [92]. They found 4.3±3.2 mm position error and 1.83°±1.77° orientation error for a 58 mm side length Apriltag marker in a 0.8-1.2 m distance at 1280×720 camera resolution. Given the identity, position and orientation of the AR markers, the coordinates of the virtual anatomical points are calculated for each frame using homogeneous coordinate transformations as described in [80] and demonstrated with source code in the Appendix. In the moment of calibration, the end point coordinates of the calibration pointer are taken as the coordinates of the calibrated anatomical point in the local coordinate system of the corresponding body segment determined by the AR marker (Figure 4.3).

4.2.2 Validation of the system

4.2.2.1 Subjects

Ten subjects of varying age (ranging from 18-84 years) participated in the study (age: 28.6±19.6 years, height: 1.71±0.06 m, weight: 66.8±17.8 kg). All participants were free from any musculoskeletal disorders. A written consent was given by the subjects after all necessary information about the procedure was presented. The study was approved by the National Science and Research Ethics Committee (21/2015).

4.2.2.2 Reference system

An 18 camera OptiTrack Flex13 motion capture camera system (Natural Point, OR, USA) was used to simultaneously track the AR markers at matching sampling frequency with the video recording set to 50 Hz. Three infra reflexive motion capture markers were fixed on each AR marker defining a trackable rigid body in the Motive software (version: 1.10.2, Natural Point, OR, USA). Coordinate systems of the rigid
bodies in Motive were aligned with the AR marker coordinate systems as it is identified by the Apriltag algorithm. This enabled the use of the same anatomical calibration on AR marker position and orientation, as well as the performance of the whole data processing described above on the same motion recorded by the two different systems. The only source of the deviations in the final gait parameters calculated by both systems is the tracking inaccuracies of the proposed solution that wanted to be identified, and possible inaccuracies of the action camera placement if the optical axis of the camera and the axis x of the motion capture system are not completely parallel. This latter error may only influence spatial gait parameters when only designated projections of anatomical points are used in the calculation (e.g. only the x coordinate of foot markers is used to calculate step size). The same applies to conventional motion capture systems when the patient’s trajectory or the placement of the treadmill is not completely parallel to the motion capture reference frame. This error is neglected in the comparison of the gait parameters but addressed in the anatomical point accuracy comparison.

4.2.2.3 Measurement procedure

Every subject performed normal walking on a treadmill moving at rates of 2.0, 3.0 and 4.5 km/h for one minute measurement intervals. There was an about one minute pause between the subjects’ trials while the recording was saved and the next capture was prepared. Recording started after the subject’s gait pattern stabilized on the treadmill (usually after 5-10 seconds of stepping on the moving treadmill). The whole procedure was repeated with and without shoes.

4.2.2.4 Accuracy of the virtual anatomical points

A marker set described in [101] – but without the heel markers – was used for virtual anatomical points (Figure 4.3). To measure the accuracy of the AR marker based system on the virtual anatomical points, an absolute comparison is required on their coordinates to the coordinates measured by the OptiTrack system. This requires a common reference system for the two measurements. Although the GoPro camera was placed with the optical axis parallel to the x axis of the OptiTrack system, this solution might not be completely accurate as discussed above. Furthermore, due to the closed structure of the camera, the exact location of the camera sensor - which is the center of the AR marker coordinate system - is difficult to align with the OptiTrack coordinate
Another issue is the time synchronization of the data. As the two systems are not integrated, neither the shutter of the cameras nor the starting of the recording are synchronized. The previous will include a uniformly distributed error as much as the reciprocal of the sampling frequency. The later error – synchronization of the starting frames – can be eliminated similarly as the data is separated into gait cycles [102] by finding key frames in both datasets based on relative marker coordinates (peaks in the difference signal of a hip and an ankle virtual point coordinate). In order to move the virtual anatomical point coordinates measured by both systems in a common reference frame, the following data manipulation was performed:

- The starting time was synchronized by removing the beginning of both recordings before the starting frame of the fifth gait cycle of the right leg.
- Based on the new common first frame, the gravity of both point clouds were moved to zero.
- Based on the common first frame angular errors of the coordinate system axis, it was corrected in the AR marker measurements to match the coordinate system of the OptiTrack system. For this purpose, coordinate root mean square error optimization was performed to identify angular errors of the AR marker system.
- The gravity shifting (zero centering) transformation defined by the first frames was applied to the whole datasets. The angular correction based on the first frame of the AR dataset was applied for the whole AR dataset.

The first 500 frames of each virtual anatomical point coordinate in each measurement were concatenated for both systems grouped, by coordinate directions and the walking speed of the measurements. The accumulated OptiTrack and AR marker based data are finally compared using Bland-Altman analysis.

4.2.2.5 Calculation of gait parameters

The exported .trk file with the marker trajectories is used by the OpenSim program to run inverse kinematics on a musculoskeletal model (Gait2354). For each time step of recorded motion data, OpenSim computes a set of joint angles that put the model in a configuration that "best matches" the experimental kinematics. This "best match" is determined by solving a weighted least squares optimization problem with the goal of minimizing marker error. Marker error is defined as the distance between an experimental marker (virtual anatomical points in our terms) and the corresponding
model marker placed on the OpenSim model anatomical points. The explanation of the joint angle calculation is summarized in the original paper of Delp et al. [103]. Contiguous motion is separated into gait cycles similarly to the method described in [102] at the peaks of coordinate x differences in the forward direction of the anterior superior iliac spine and the medial ankle. All compared parameters average values for a test case of the subjects’ gait cycles. The calculated spatial and angular parameters are described in Table 4.2. The range of motion (ROM) is defined for the angular gait variables, as the difference of the maximum and minimum values of the joint trajectory. The processing of gait cycles, OpenSim joint data and spatiotemporal gait parameters are calculated by a custom Matlab script in Matlab version R2017b (MathWorks, Natick, Massachusetts, USA).

**Table 4.2. Calculated gait parameters**

<table>
<thead>
<tr>
<th>Parameter name / dimension</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride length [m]</td>
<td>Distance by which each foot is in front of the other one at heel strike. Measured by medial ankle coordinates.</td>
</tr>
<tr>
<td>Step length [m]</td>
<td>Distance by which the foot moves forward in one gait cycle. Measured by medial ankle coordinates.</td>
</tr>
<tr>
<td>Walking base [m]</td>
<td>The side to side distance between the line of the two feet. Measured by medial ankle coordinates.</td>
</tr>
<tr>
<td>Cadence [steps/minute]</td>
<td>The total number of gait cycles taken within a minute. Calculated from the average cycle time of the individual gait cycles.</td>
</tr>
<tr>
<td>Hip flexion ROM [°]</td>
<td>Range of motion (difference of the maximum and minimum values of the joint angle trajectories) of the angular parameters averaged for the gait cycles of the trial as calculated by the OpenSim model described in [103].</td>
</tr>
<tr>
<td>Hip addiction ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Hip rotation ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Knee angle ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Pelvis tilt ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Pelvis list ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Pelvis rotation ROM [°]</td>
<td></td>
</tr>
<tr>
<td>Pelvis tx ROM [m]</td>
<td>Range of translational motion (difference of the maximum and minimum coordinates) of the pelvis center coordinates averaged for the gait cycles of the trial as calculated by the OpenSim model described in [103].</td>
</tr>
<tr>
<td>Pelvis ty ROM [m]</td>
<td></td>
</tr>
<tr>
<td>Pelvis tz ROM [m]</td>
<td></td>
</tr>
</tbody>
</table>

The studied values are the mean values of the multiple gait cycles for each trials ROM: range of motion
4.2.2.6 Statistical analysis

The calculated gait parameters were compared between the measurement systems. As each parameter was calculated for each trial and data recording was simultaneous on the two systems, the datasets could be paired. Root mean square errors (RMSE) for each averaged parameter were calculated between the datasets to characterize the accuracy of the AR marker system. Additionally, a Bland-Altman analysis [104] was conducted on these datasets to characterize correlation, limits of agreement on a 95% confidence interval, mean error and a reproducibility coefficient (RPC=1.96SD) between the measurement systems. Additionally minimal detectable change (MDC) was calculated from the within-subject gait variability for both systems as:

\[ MDC = SEM \times 1.962 \times \sqrt{2} \]  \hspace{1cm} (4.1)

where SEM is the standard error of measurement calculated as

\[ SEM = \sqrt{\frac{\sum_{i=0}^{n} SD_{i}^2}{n}} \]  \hspace{1cm} (4.2)

where \( i \) iterates over the measurements, \( n \) is the number of measurements, and \( SD \) is the standard deviation of the gait parameters for the individual gait cycles within the \( i \)-th measurement. While SEM is frequently calculated from the intraclass correlation coefficient, Baker [105] recommends the above simple method for calculating SEM to describe within-subject variability in gait analysis. The accuracy of anatomical landmark placements [80] was not studied, as single anatomical calibrations were used for both measurement system.

4.3 Results

4.3.1 Sample size

The measurements of two subjects had to be excluded from the study later on due to improper marker placement which was realized during the evaluation of the results. The elderly subject failed to perform the 4.5 km/h trials. The final number of trials therefore is \( n=46 \) which includes trials from eight subjects with and without shoes.
at 2.0, 3.0 and 4.5 km/h walking speeds, except the elderly subject where only 2.0 and 3.0 km/h trials were performed. This sample size produces 0.25SD standard error in the evaluation of the limit of agreement values in the Bland Altman analyses (1.71SD/√n according to Bland and Altman [104]).

4.3.2 Accuracy of virtual anatomical points

The summarized results of the Bland-Altman analysis for the virtual anatomical point position comparison are shown in Table 4.3. The results can be analyzed separately in each of the three coordinates (see directions on Figure 4.3) and walking speeds (2, 3, 4.5 km/h). The averaged slope of the regression lines was 1.05, 1.0 and 1.02 in directions x, y, and z, respectively, while the averaged r² value was 0.98, 1.0, and 0.99, respectively. Due to the data manipulation to align the two reference frames, no significant bias can be observed on any coordinates. RPC values are the largest in direction x (mean: 33.92, SD: 3.42).

Table 4.3. Results of the Bland-Altman analysis on coordinates of virtual anatomical landmarks

<table>
<thead>
<tr>
<th>Anatomical landmark coordinates</th>
<th>r²</th>
<th>Slope</th>
<th>RPC (mm)</th>
<th>Mean error (mm)</th>
<th>95% confidence interval* (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2 km/h</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0.98</td>
<td>1.06</td>
<td>32.6</td>
<td>-0.05</td>
<td>(-33, 33)</td>
</tr>
<tr>
<td>y</td>
<td>1</td>
<td>1</td>
<td>24.24</td>
<td>-1.6</td>
<td>(-26, 23)</td>
</tr>
<tr>
<td>z</td>
<td>0.99</td>
<td>1.02</td>
<td>13.97</td>
<td>0.61</td>
<td>(-13, 15)</td>
</tr>
<tr>
<td>3 km/h</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0.98</td>
<td>1.05</td>
<td>31.35</td>
<td>1</td>
<td>(-30, 32)</td>
</tr>
<tr>
<td>y</td>
<td>1</td>
<td>1</td>
<td>26.42</td>
<td>-0.03</td>
<td>(-26, 26)</td>
</tr>
<tr>
<td>z</td>
<td>1</td>
<td>1.02</td>
<td>11.97</td>
<td>1.5</td>
<td>(-11, 13)</td>
</tr>
<tr>
<td>4.5 km/h</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0.98</td>
<td>1.05</td>
<td>37.8</td>
<td>1.6</td>
<td>(-36, 39)</td>
</tr>
<tr>
<td>y</td>
<td>1</td>
<td>1</td>
<td>28.82</td>
<td>-0.2</td>
<td>(-29, 29)</td>
</tr>
<tr>
<td>z</td>
<td>0.99</td>
<td>1.02</td>
<td>14.77</td>
<td>2.6</td>
<td>(-12, 17)</td>
</tr>
<tr>
<td>Mean (SD)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0.98 (0)</td>
<td>1.05 (0.01)</td>
<td>33.92 (3.42)</td>
<td>0.85 (0.84)</td>
<td></td>
</tr>
<tr>
<td>y</td>
<td>1 (0)</td>
<td>1 (0)</td>
<td>26.49 (2.29)</td>
<td>-0.61 (0.86)</td>
<td></td>
</tr>
<tr>
<td>z</td>
<td>0.99 (0.01)</td>
<td>1.02 (0)</td>
<td>13.57 (1.44)</td>
<td>1.57 (1.0)</td>
<td></td>
</tr>
</tbody>
</table>

* 95% confidence interval equals the range of the bias ± 1.96 times the standard deviation of the differences. It is also referred to as the limit of agreement.
4.3.3 Deviation of gait parameters

Detailed results of the Bland-Altman analysis, RMSE and MDC values are presented in Table 4.4 for each calculated gait parameter. Overall, distance type parameters showed RMSE as smaller than or equal to 23 mm with a mean error (bias) smaller than or equal to 7 mm and CV smaller than 4.6%. The mean error of the angle type ROM parameters indicated significant deviations between the two measurement systems with a mean error range between 1.27° and 3.91° and an RMSE range between 2.55° and 6.72°. The detection of the pelvis position range of motion also showed small mean errors (≤ 1 mm) with RMSE between 5 and 8 mm, but larger CV (16.5-19.4%). The only time based parameter of cadence (step frequency) yielded a mean error of 0.18 steps/minute with RMSE of 1.116 steps/minute and 1.19% CV. Most RMS errors are in the range of MDC of the OptiTrack system. In case of hip adduction and rotation ROM and pelvis list and rotation angles the RMS error is larger than the MDC values. For illustrating the differences between the measurement systems, joint angle trajectories of a subject in the 2 km/h trial are shown in Figure 4.4, where it is visible that mean differences in certain parameters are in the range of gait variability (hip rotation, pelvis list and positions), while other parameters have significant offset errors (pelvis rotation and tilt, hip flexion and addiction). Joint trajectories measured by the AR marker based system (red) are drawn on top of the trajectories measured by the OptiTrack (blue). The dashed lines are the averaged joint trajectories during the trial, while the band around them is the ± intrasubject standard deviation at each percent of the gait cycle 95% confidence interval representing the gait variability. Differences of the two mean trajectories (black) are also illustrated in the figures. Due to camera position offset, the pelvis tx, ty and tz position parameters are zero centered for easier comparison. The range of motion gait parameters are defined by the difference of the maximum and minimum values of the averaged joint trajectories.
Figure 4.4. Comparison of joint angle trajectories.
Table 4.4. RMS error and Bland-Altman analysis of gait parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>RMSE</th>
<th>$r^2$</th>
<th>Slope</th>
<th>RPC</th>
<th>CV (%)</th>
<th>Mean error</th>
<th>95% confidence interval of error</th>
<th>MDC (Opti-Track)</th>
<th>MDC (AR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride length [m]</td>
<td>0.013</td>
<td>0.996</td>
<td>0.988</td>
<td>0.026</td>
<td>1.201</td>
<td>-0.002</td>
<td>(-0.028; 0.024)</td>
<td>0.059</td>
<td>0.052</td>
</tr>
<tr>
<td>Step length [m]</td>
<td>0.023</td>
<td>0.956</td>
<td>0.996</td>
<td>0.044</td>
<td>4.105</td>
<td>-0.007</td>
<td>(-0.050; 0.037)</td>
<td>0.060</td>
<td>0.066</td>
</tr>
<tr>
<td>Walking base [m]</td>
<td>0.023</td>
<td>0.915</td>
<td>0.947</td>
<td>0.016</td>
<td>4.594</td>
<td>-0.003*</td>
<td>(-0.019; 0.013)</td>
<td>0.049</td>
<td>0.040</td>
</tr>
<tr>
<td>Hip flexion ROM [°]</td>
<td>4.666</td>
<td>0.848</td>
<td>0.898</td>
<td>8.848</td>
<td>12.081</td>
<td>1.274*</td>
<td>(-7.574; 10.122)</td>
<td>6.725</td>
<td>6.629</td>
</tr>
<tr>
<td>Hip addiction ROM [°]</td>
<td>3.489</td>
<td>0.647</td>
<td>0.764</td>
<td>5.129</td>
<td>17.306</td>
<td>2.324*</td>
<td>(-2.805; 7.452)</td>
<td>3.398</td>
<td>3.442</td>
</tr>
<tr>
<td>Hip rotation ROM [°]</td>
<td>6.728</td>
<td>0.459</td>
<td>0.687</td>
<td>10.792</td>
<td>34.586</td>
<td>3.910*</td>
<td>(-6.882; 14.701)</td>
<td>3.699</td>
<td>3.509</td>
</tr>
<tr>
<td>Knee angle ROM [°]</td>
<td>3.607</td>
<td>0.945</td>
<td>1.040</td>
<td>6.555</td>
<td>5.809</td>
<td>1.396*</td>
<td>(-5.159; 7.952)</td>
<td>4.283</td>
<td>4.487</td>
</tr>
<tr>
<td>Pelvis tilt ROM [°]</td>
<td>2.554</td>
<td>0.558</td>
<td>1.128</td>
<td>4.388</td>
<td>34.497</td>
<td>1.272*</td>
<td>(-3.116; 5.660)</td>
<td>3.779</td>
<td>4.179</td>
</tr>
<tr>
<td>Pelvis list ROM [°]</td>
<td>3.750</td>
<td>0.297</td>
<td>0.518</td>
<td>6.761</td>
<td>33.792</td>
<td>1.556*</td>
<td>(-5.205; 8.316)</td>
<td>2.432</td>
<td>4.067</td>
</tr>
<tr>
<td>Pelvis rotation ROM [°]</td>
<td>3.678</td>
<td>0.641</td>
<td>0.887</td>
<td>6.004</td>
<td>31.249</td>
<td>-2.086*</td>
<td>(-8.090; 3.918)</td>
<td>3.487</td>
<td>4.922</td>
</tr>
<tr>
<td>Pelvis tx ROM [m]</td>
<td>0.008</td>
<td>0.573</td>
<td>0.701</td>
<td>0.016</td>
<td>19.483</td>
<td>&lt;0.001</td>
<td>(-0.015; 0.016)</td>
<td>0.021</td>
<td>0.023</td>
</tr>
<tr>
<td>Pelvis ty ROM [m]</td>
<td>0.005</td>
<td>0.888</td>
<td>1.243</td>
<td>0.010</td>
<td>16.159</td>
<td>0.001</td>
<td>(-0.008; 0.011)</td>
<td>0.008</td>
<td>0.009</td>
</tr>
<tr>
<td>Pelvis tz ROM [m]</td>
<td>0.008</td>
<td>0.713</td>
<td>0.813</td>
<td>0.016</td>
<td>16.585</td>
<td>&lt;0.001</td>
<td>(-0.016; 0.016)</td>
<td>0.019</td>
<td>0.020</td>
</tr>
<tr>
<td>Cadence [steps/minute]</td>
<td>1.116</td>
<td>0.995</td>
<td>1.010</td>
<td>2.180</td>
<td>1.191</td>
<td>0.187</td>
<td>(-1.993; 2.368)</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

*Significant mean error (p<0.05), ROM: range of motion, RMSE: root mean square error, $r^2$: squared Pearson r-value of the correlation plot, Slope: the slope RPC: reproducibility coefficient (1.96*SD), CV: coefficient of variation (SD of mean values in %)

4.4 Discussion

The present chapter described the concept and possible technical solutions of a gait analysis system using a single action camera and AR markers, and its aim was to validate an exemplified implementation of the concept with simultaneous measurements of a conventional motion capture system. A validation of the system has been
performed by an 18 camera OptiTrack motion capture system on healthy gait at different walking speeds (2.0, 3.0 and 4.5 km/h) on a treadmill and by comparing 3D anatomical point coordinates (Table 4.3) and several gait parameters (Table 4.2, Table 4.4) using both systems. Generally, significant mean errors of angular ROM gait parameters (marked with * in Table 4.4) can be observed in the AR marker system; however, the errors of the distance type parameters are relatively smaller, except for the walking base.

Studying the absolute errors of virtual anatomical points (Table 4.3), it is obvious that absolute coordinate errors depend on the direction (Figure 4.3), as the reproducibility coefficients (RPC) - especially in the direction of motion (x) and also in the upward (y) direction - are larger than errors in the medial-lateral (z) direction. These errors are also larger at higher walking speed where a larger range of motion can be observed (Table 4.3). These errors can be traced back to the calculation of marker orientations. The coordinate system convention for understanding this explanation is shown in Figure 4.3 as axis x points in the forward direction of the movement, y points upward, and z points to the right. The orientation of the markers with respect to the camera is calculated through homography transformation from the pixel coordinates of the marker corners [99]. In the applied orientations of the markers, the sideways accuracy of virtual anatomical points is mostly influenced by the accuracy of marker rotation around axis x and the marker position in direction z, which can be highly accurately calculated from the high resolution image of a properly calibrated camera. On the other hand, marker rotations around axes y and z and the x coordinate calculation of the marker are more influenced by camera perspective due to homography transformations.

On the other hand, the angular errors of the AR marker detection affect more the observed joint angles (e.g. knee angle and pelvis flexion) and the virtual ankle x coordinates that are used to calculate step length and stride length (Table 4.3). This suggests that the location of the camera may affect the recording quality, and these joint angles might be more accurate when the gait analysis is done in the sagittal plane (the camera and AR markers are on the side of the subject). In this setup only one leg can be analyzed. Preliminary results with the same measurement setup showed that unfortunately the results are not significantly better this way [106], as all the markers
cannot face the camera continuously and even in stationary position the markers do not line in a plane which is perpendicular to the optical axis and the orientation estimation is also introduce errors in this setup. The 3D motion capture is performed regardless of the camera position and joint angles can be captured in each plane of the motion.

The between-system differences for some parameters are in the range of differences between stereophotogrammetric motion capture systems such as the one used as reference in the present research (the OptiTrack camera system). Thewlis et al. found $2.7^\circ$ mean differences in knee range of motion between two commercial motion capture camera systems in a simultaneously measured gait trial [66]. This is comparable to our even smaller deviation in knee range of motion ($1.39^\circ$). The proposed system can accurately measure human gait with maximum $3.91^\circ$ mean error (hip rotation ROM) in the studied angular parameters and maximum $7$ mm mean error in spatial parameters. On the other hand, higher RMSE values show us that the deviations are nondeterministic, therefore longer or multiple averaged measurements are required for calculating averaged gait parameters for walking trials.

The calculated MDC values of both system are of similar values to those published in the literature (e.g. in Fernandes et al. [107] stride length: $0.09$ m, step length: $0.05$ m, step width: $0.02$ m, and peak parameters for hip flexion $7.9^\circ$, hip adduction $3.9^\circ$, knee angle $5.5$-$7^\circ$, or in Bates et al. [108] ROM parameters for hip flexion: $2.51^\circ$, hip adduction $1.48^\circ$, hip rotation: $4.35^\circ$, knee angle: $5.34^\circ$, pelvis tilt: $1.34^\circ$, pelvis rotation: $1.88^\circ$). Most RMS errors are in the range of MDC of the OptiTrack system, thus this difference is not statistically significant when deviations from a healthy reference group is sought. In case of hip adduction and rotation, pelvis list and rotation angles the RMS error is larger than the MDC values and even in the pelvis angles MDC is slightly ($0.4$-$1.6$ degree) larger than in the OptiTrack system. In this case repeated measurements between the two systems on the same patient would show deviations. On the other hand other MDC values by the two system are very similar (Table 4.4).

Compared to gait analysis reports from the literature [82, 109–111], within-subject and inter-subject differences could be shown with these errors by the proposed system in common gait analysis applications such as the following examples. Kim and Eng [109] have studied inter-subject angular differences in the paretic and non-paretic
legs of stroke survivors. For the knee flexion ROM they found 16.1° mean difference between legs [109], while the RMS error of the knee flexion ROM is 3.6° between the AR marker system and OptiTrack, thus this deviation could have been shown by the proposed system. Bejek et al. have studied the effect of walking speed on gait parameters in patients with osteoarthritis and healthy controls [82]. In their study, step size differs by more than 200 mm in each group between different walking speeds (1, 2, 3 and 4 km/h), while the RMS error of this parameter of our system is only 23 mm. The asymmetry of step length in the osteoarthritis group is between 38.1 and 217 mm for the different walking speeds. Similarly, the asymmetry of the knee angle ranges between 6.4° and 15.9° at 1 and 4 km/h walking speed [82], while the RMS of knee angle is 3.6 with our system. Derrick [110] collected knee angle measurements from the literature at foot contact with different experimental procedures: knee contact angle changes between 10% understride and 10% overstride by 2°, due to fatigue by 4.4°, between smooth and irregular walking surface by 1.5°, and between short and long grass on the walking surface by 4.2°. Duffel and Jordan found an insignificant 2.5° difference in the largest knee angle between 18-30 and 60+ year old healthy subjects [111]. The above examples demonstrate multiple use cases of gait analysis where the present gait parameters are used with smaller (mostly statistically not significant) or larger differences between cases compared to the measurement errors of our system. The very small differences cannot be reliably measured with the present exemplified system, but significant deviations could be observed with even the present rudimentary implementation of an AR marker based motion capture. At this point it is important to consider that Thewlis et al found 2.7° mean difference of the knee ROM between commercially available multi camera motion capture systems [66]. The usability of the present (and any) system depends on the expected effect size of research.

Although the exemplified implementation of the proposed approach does not yet fulfil all requirements expected from the high-end user friendly multi-camera motion capture system, the proposed approach can be utilized in the development of a consumer grade low cost motion capture system by refining some of the technological cornerstones, e.g. more precise camera calibration, more precise AR marker tracking and real-time behavior. The most important upgrade could be the improvement of marker orientation precision. One solution could be the usage of a different augmented
reality marker using a microlense array which promises higher orientation accuracy [112].

The study introduced a new, mobile and affordable gait analysis approach using augmented reality markers fixed on body segments recorded by an action camera. The solution was introduced and validated using an OptiTrack motion capture system with multiple walking speeds and subjects. The proposed method shows some differences in the raw coordinates of virtually tracked anatomical landmarks (RPC 33.92 mm in direction $x$, 26.49 in direction $y$ and 13.57 mm in direction $z$) and gait parameters compared to the reference system (Table 4.4); however, these differences are comparable to previously reported differences between commercial motion capture systems. Accuracy might be improved by more advanced AR marker tracking.

3. Contribution

Motion tracking of body segments as rigid bodies can be performed with the image processing methods of augmented reality markers that determines position and orientation which can be calculated from the image of a single camera.

Related publications: [P9-10]
5 MAIN CONTRIBUTIONS

This chapter summarizes the dissertation in the form of theses along with short background explanation and practical application.

1. Contribution

The following parameters should be used to describe the motion of the center of pressure and deviations of the balancing capability for the purpose of stability tests:

- the path length of the foot center of pressure,
- largest anterior and posterior (forward and backward) deviation from the average center of pressure (Figure 5.1),
- the largest continuous amplitudes projected onto the axes (Figure 5.2),
- load distribution difference between lower limbs.

Figure 5.1. CoP AP+ ($d_1$) and AP- maximum deviation ($d_2$) and angles ($\alpha$, $\beta$)
Figure 5.2. CoP largest amplitude

Related publications: [P1-6]

Background

From the pressure distribution of the foot center of pressure and its movement can be calculated, which characterizes the standing balance (static balancing capability under quiet standing). Foot center of pressure can be recorded using pressure distribution measuring plates.

In the literature numerous center of pressure based stabilometry parameter are utilized, but many of them characterizes the same principle in different notation. An obvious example for this is the path length, fixed sampling time and average velocity which linearly related. This makes the comparison of results of different studies complicated. The contribution is based on the reliability analysis of the parameter set that was narrowed by correlation and variance analysis. Among numerous center of pressure parameters collected form the literature the independent parameters can be selected using correlation analysis [P1, P2]. The sensitivity of these parameters were evaluated using variance analysis in stance types of reportedly different balancing characteristics (bipedal stances with eyes open and closed, single leg stance on the dominant leg with eyes open) [P1, P2]. Those parameters which did not indicate significant or not significant difference between the stance types were excluded from future analysis. The reliability of the narrowed parameter list is characterized by the
The intraclass correlation coefficient (ICC), the standard error of the measurement (SEM) and the coefficient of variation (CV) calculated for repeated measurements [P3]. The result of the analysis is a proposal made for the reliable and independent parameters for balancing capability evaluation that can be calculated from the motion of the foot center of pressure.

**Practical application**

The narrowed center of pressure parameter list was applied in the following studies:

- study the effect of bad posture on standing balance in school aged children [P5],
- study the effect of pes planus on standing balance in school aged children [P6],
- study the effect of bilateral knee osteoarthritis on standing balance [P4].

The proposed parameters can be used regardless of age to study deviations from healthy state.

**2. Contribution**

The static error of stereo-photogrammetry based motion capture systems can be determined using engineering surveying methods, whereby scaling error can be defined by simultaneous measurements of spatial position (different heights) of widely spaced markers in a common base coordinate system with the motion capture and a surveying systems, thus the system can be calibrated.

Related publications: [P7-8]

**Background**

The stereo-photogrammetry based motion capture systems determine the spatial position of markers based on triangulation. To achieve this the relative position of the cameras and a reference coordinate system has to be known which is identified by extrinsic camera calibration in the daily routine using a marker equipped calibration wand with known geometry. When the distances of the calibration markers deviates from the theoretical distance used by the calibration software, then scaling error occurs
in the calibration which leads to consequential marker coordinate errors that correlates with distance from the origin. The uncertainty of a simple measurement tool (caliper) used for measuring the calibration wand marker distances can yield marker coordinate errors of the range of centimeters in several meters away from the origin in a calibrated motion capture system which has theoretical submillimeter accuracy. Common methods to evaluate motion capture system accuracy are based on the measurement of small relative motions of markers or the measurement of distance of rigidly fixed markers moved together in the measurement volume. These methods can not detect scaling errors as it affects both the moved marker coordinate and the reference position or both marker coordinates. The scaling error can only be detected using absolute position measurements e.g. engineering surveying. The marker coordinates should be measured simultaneously with the motion capture system and surveying station in a common reference frame. Comparing the measured coordinates the scaling error can be determined. The scaling error can be also determined by measuring the distance of markers placed in large distance inside the measurement volume using the motion capture system and a validated tape measure. With this method the accuracy of scale error determination improves with larger marker distances.

**Practical application**

In case of an OptiTrack Flex13 motion capture system (NaturalPoint, Oregon, USA) with 18 cameras a scaling factor of 0.9992 was observed by comparing the own measurements of the system to surveying measurements of marker coordinates. This resulted 4 mm static position error 4 meters from the origin. This error increases to 8 mm error in 10 m distance from the origin. With an identified scaling factor the error of the marker coordinates can be compensated (Figure 5.3). Compensation not only required for biomechanical measurements, but also in robotic applications, as the outstanding absolute accuracy is required compared to robot odometry.
Figure 5.3. 3D error vector in the original (A) and the optimally downscaled (B) condition. (Error vector magnification was set to 100 for better visibility.)

3. Contribution

Motion tracking of body segments as rigid bodies can be performed with the image processing methods of augmented reality markers that determines position and orientation which can be calculated from the image of a single camera.

Related publications: [P9-10]

Background

Optical motion capture systems are used to record spatial coordinates of anatomical landmarks in gait analysis. The spatial coordinates of anatomical landmarks are mapped to the coordinates of marker attached to them. When marker clusters are used on the body segments to minimize soft tissue artefact, during an anatomical calibration the location (position vector) of the individual anatomical landmarks is determined in the local coordinate system of the marker clusters. The position and orientation of the marker cluster is measured by means of stereo-photogrammetry, which characterizes the motion of body segments as rigid bodies. The position of anatomical landmarks is calculated from the position and orientation of the clusters and the position vectors.
During the implementation of augmented reality – similarly to the marker clusters of optical motion capture systems – the position and orientation of the augmented reality markers are calculated relative to the camera. The spatial coordinates of the calibrated anatomical landmarks can be calculated from these data and the position vector relative to the augmented reality markers. Consequently the technical implementation between the two fields are interchangeable and augmented reality technology can be used for biomechanical purpose motion capture [P9-10]. In this case the motion of anatomical landmarks can be tracked by augmented reality markers attached to body segments (Figure 5.4).

**Figure 5.4.** Calibration of anatomical points using the calibration pointer and augmented reality markers that represent position and orientation of body segments
**Practical application**

Augmented reality based motion capture can be a cheap alternative of the high end optical motion capture systems. A criteria of the analysis is that every marker should be seen from one direction, thus the system is capable for treadmill trials with fixed camera placement [P9-10]. To calculate joint angles only the relative position and orientation of the markers are required, thus a measurement can be performed from a person walking on the ground followed by a smartphone camera.
AUTHOR’S PUBLICATIONS RELATED TO THE CONTRIBUTIONS


BIBLIOGRAPHY


D. Thewlis, C. Bishop, N. Daniell, G. Paul, Next-generation low-cost motion


6 MAIN CONTRIBUTIONS IN HUNGARIAN / TÉZISEK MAGYARUL

Jelen melléklet az 5. fejezet fordítása, mely magyarul is összefoglalja a téziseket, és a hozzájuk tartozó rövid magyarázatokat, alkalmazásokat.

1. tézis

Az egyensúlyozó képesség változásait az állás közben mért talpnyomás eloszlásból meghatározott nyomás középpont (CoP) pályagörbéjéből számított következő paraméterekkel célszerű leírni:

- a talpnyomás középpont bejárt útjának hossza egységes mérési idő alatt
- átlagos talpnyomás középponthoz képesti legnagyobb anterior és posterior (előre-hátra) irányú eltérés (6.1. ábra)
- a legnagyobb, tengelyekre vetítve egybefüggő kilendülések amplitúdója (6.2. ábra)
- az alsó végtagok közötti terhelés százalékos megoszlásának különbsége.

![Diagram](image)

6.1.ábra. CoP AP+ \( (d_1) \) és AP- legnagyobb kitérések \( (d_2) \) és szögeik \( (\alpha, \beta) \)
6.2.ábra. CoP bejárt út legnagyobb amplitúdója

Kapcsolódó publikációk: [P1-6]

Háttér
Talpmomás eloszlás mérő lapal a nyomáloszlás állás közben rögzíthető. A
nyomáloszlásból a nyomásközéppont és annak változása számítható, amely az
állásstabilitást (statikus, állás közbeni egyensúlyozó képesség) jellemzi.

A szakirodalomban számos talpmomás középpont mozgásán alapuló állásstabilitási
paramétert alkalmaznak, de több paraméter ugyanazt a jellemzőt írja le más
megfogalmazásban. Erre legnyilvánvalóbb példa a bejárt út hossza adott mérési idő alatt,
és átlagsebesség, amelyek lineáris kapcsolatban állnak. Az irodalomban található sok
paraméter megnehezíti a különböző tanulmányok eredményeinek összehasonlítását. A
tézis alapja a korreláció és varianciaanalízissel előszürt paraméterek megbízhatósági
vizsgálattal megállapított ismétlési pontosság vizsgálata. Az irodalomból összegyűjtött
paraméterek közül korreláció vizsgálattal a független távolság-idő alapú- és a frekvencia
vizsgálaton alapuló paraméterek kiválaszthatók [P1, P2]. Ezek érzékenységét
varianciaanalízissel vizsgáltuk bizonyítottan eltérő biomechanikai tulajdonságú
állástípusok során (nyitott és csukott szemes kétlábas állás, nyitott szemes félábas állás)
[P1, P2]. A változást nem vagy csak bizonytalanul érzékelni képes paramétereket a
további vizsgálatból kizártuk. A szűkített listán végzett megbízhatósági vizsgálat
egymás után többször megismételt mérésekből számított paraméterekre az osztályon
belüli korrelációs együtthatót (ICC - intraclass correlation coefficient), a mérés standard hibáját (SEM – standard error of measurement) és a relatív szórást (CV – coefficient of variation) elemzi [P3]. A vizsgálatsorozat eredményeként az állásstabilitás jellemzésére megbizható és egymástól független paraméterek használatát javasolom.

**Gyakorlati alkalmazás**

A talpnyomás középpont mozgásán alapuló szűkített paraméterlistát az alábbi kutatásokban használtuk:

- hanyag tartás hatásának vizsgálata az iskoláskorú gyerekek állásstabilitására [P5];
- lúdtalp hatásának vizsgálata iskoláskorú gyerekek állásstabilitására [P6];
- térdizületi kopás hatásának vizsgálata az idősek állásstabilitására. [P4].

A javasolt paraméterek a vizsgált személyek életkorától függetlenül hatékonyan alkalmazhatóak akár a fiziológiai változások kimutatására.
2. tézis

Sztereo-fotogrammetrián alapuló optikai mozgásrögzítő rendszerek (motion capture) statikus hibája geodéziai mérésekkel meghatározható, melyek során báziskoordináta rendszeren térbeli (különböző magasságú) és egymástól távol elhelyezett markerek helyzetét optikai és a geodéziai mérésekkel egy időben meghatározva a skálázási hiba definiálható, és az így vizsgált rendszer kalibrálható.

Kapcsolódó publikációk: [P7-8]

Háttér magyarázat

A sztereo-fotogrammetrián alapuló mozgásrögzítő rendszerek a háromszögelés módszerével határozzák meg a markerek térbeli koordinátáit. Ehhez szükséges a kamerák egymáshoz viszonyított helyzetének és egy referencia koordináta rendszernek az ismerete, amelyet a napi gyakorlatban külső (extrinsic) kalibrációval határoznak meg, amelyhez egy markerekkel felszerelt, ismert geometriájú kalibráló marker elrendezést – kalibráló pálcát – használnak. Ha a kalibráló pálca markerei közötti távolság eltér a kalibráló programban referenciaéértéktől, akkor a kalibrációban ez skálázási hibát okoz. Ez a mért marker koordinátákból az origótól mért távolsággal arányosan növekvő koordinátát eredményez. Ezt jól mutatja, hogy az egyszerűbb mérőeszközök (pl. tolómérő) mérési hibája a kalibráló markertávolság meghatározásában, az origótól több méterre már centiméteres koordinátát kialakítja a kalibráló program, és a skálázási hiba okozhat az elvileg milliméter alatti pontosságú kalibrált rendszerben. Az irodalomban a pontosság vizsgálatát kis relatív elmozdulásokkal, vagy egymáshoz mereven rögzített markerek mérőterén vizsgálták. E módszerek a skálázási hibát nem tudják kimutatni, mert a skálázási hiba egyformán hat az elmozdult koordinátára és referencia pozícióra vagy marker koordinátákrá. A skálázási hiba csak abszolút pozíció mérések segítségével pl. geodéziai mérésekkel mutatható ki. A mérőtérben elhelyezett markerek koordinátáját közös vonatkoztatási rendszerben geodéziai mérőállomással és a mozgásrögzítő rendszerrel is egy időben kell mérni. A két módszerrel mért térbeli koordináták összehasonlításával a skálázási hiba meghatározható. A pontosságot befolyásoló skálázási hiba egymástól nagy távolsában elhelyezett markerek távolságát a vizsgált rendszerrel és hitelesített mérőszalaggal
megmérve is meghatározható. A skálázási hiba meghatározás pontossága a markerek távolságának növelésével javul.

**Gyakorlati alkalmazás**

Egy $4 \times 2,5$ m mérési alapterülettel rendelkező 18 kamerás OptiTrack Flex13 (NaturalPoint, Oregon, USA) kamerarendszer esetén a geodéziai mérés és a rendszer saját mérésének összehasonlításával 0,9992 skálázási faktort állapíttunk meg, amely 4 mm statikus hibát okoz a marker abszolút pozíciójában 5 méterre az origótól. 10 méter mérési távolság esetén ez a hiba 8 mm. A skálázási faktor ismeretében a hiba kompenzátható (6.3. ábra). A kompenzáció nem csak a biomechanikai vizsgálatok során fontos, hanem robotikai alkalmazásokban is szükséges, mivel a rendszer használata a kiemelkedő abszolút pontosságot igényli a robot odometriához képest.

6.3.ábra. Eredeti 3D hibavektorok az OptiTrack rendszer marker mérései és a geodéziai mérések között (A), valamint az optimált visszaskálázott mérő térben mért hibavektorok (B). A láthatóság érdekében a hibavektorok hossza 100-as nagyítási tényezővel van ábrázolva.
3. tétis

A sok-kamerás sztereofotogrammetria módszerekben alapuló hagyományos mozgásvizsgáló rendszerek a testszegmensek követésére merev marker csoportokat használnak, hogy az egyedi markereknél meglévő bőrmozgást minimalizálják.

A testszegmensek merev testként való mozgáskövetésére Augmented Reality markerek pozíciót és orientációt meghatározó képfeldolgozása is alkalmas, amely egy kamera képével is kivitelezhető.

Kapcsolódó publikációk: [P9-10]

Háttér magyarázat

Optikai mozgásrögzítő rendszerekkel a mozgásjellemzők számításához szükséges anatómiai pontok térbeli koordinátáit rögzítjük. Az anatómiai pontok térbeli helyzete megegyezik a rájuk közvetlenül rögzített markerek térbeli helyzetével. A bőrmozgás csökkentésére használt marker csoportok használata során az ügynevezett kalibrációs fázisban az anatómiai pont térbeli helyzetét (helyvektorát) a marker csoport által meghatározott lokális koordináta rendszerben kell meghatározni. A marker csoport térbeli helyzete sztereofotogrammetria módszerekkel meghatározható, amely a merev testként jellemzett testszegmens mozgását írja le. Az anatómiai pont térbeli helyzete a lokális koordináta rendszerbeli helyvektor segítségével a marker csoport orientációjából és pozíciójából számolható.

Augmented Reality (AR - kiterjesztett valóság) megvalósítása során - az optikai mozgásjellemző rendszer markercsoportjaihoz hasonlóan - az AR markernek a kamerához viszonyított pozíciójából és orientációjából a kalibrációs fázisban kijelölt anatómiai pontok térbeli helyzete számítható (6.4. ábra). Ebből következően a technikai megvalósítás a két terület között felcserélhető, és AR marker követési technológiával is végezhető mozgásvizsgálat [P9-10]. Ebben az esetben a vizsgált személy testszegmenseire AR markereket rögzítve követhető az anatómiai pontok mozgása.
6.4.ábra. Anatómiai pont kalibráció AR markerekkel végzett mozgáskövetés során, ahol a testszegmensek pozícióját és orientációját az AR markerek segítségével követik.

Gyakorlati alkalmazás

AR markerrel végzett mozgásvizsgálat az optikai mozgásrögzítő rendszerekhez képest olcsó alternatívát jelenthet. A vizsgálat feltétele, hogy a markereknek egy irányból mindig láthatónak kell lenniük, így fix kameralrendezés futópadon történt mozgásvizsgálatára alkalmas [P9-10]. Az ízületi szögek meghatározásához a markerek relatív helyzetét kell csak rögzíteni, így a talajon járó személyt mobiltelefon kalibrált kamerájával követve is végezhető a mérés.
7 APPENDIX

7.1 C# implementation of anatomical point calculation and calibration

This supplementary material explains the process of anatomical calibration and virtual anatomical point tracking using augmented reality markers. First the relevant data classes are introduced, then the implementation of necessary mathematical formulas for homogeneous coordinate transformation (matrix rotations). Then the process of anatomical calibration is explained where the local coordinates of the virtual anatomical points are set with respect to the actually tracked body segment coordinates and orientations through the augmented reality marker tracking.

The Segment class represents a body segment defined by an AR marker position and orientation provided by any augmented reality framework, e.g. AprilTag AR marker tracking (downloadable from: https://april.eecs.umich.edu/software/apriltag)

```csharp
public class Segment
{
    public String name;

    public Vector<double> position;
    public Vector<double> orientation; //ZXZ rotational euler conventions

    public bool isVisible;

    public Segment(String name)
    {
        this.name = name;

        position = Vector<double>.Build.Dense(3);
        orientation = Vector<double>.Build.Dense(3);

        isVisible = false;
    }
}
```

The Marker class represents a virtual anatomical point that belongs to a certain body segment through its segmentID variable which refer to the index of the corresponding segment in an array of body segments that defines our biomechanical model.

The local coordinates refer to the anatomical point coordinates respect to the AR marker centered moving reference frame attached to the corresponding body segment where the anatomical point is defined and moves rigidly together with the segment.
The global coordinates of the anatomical points refer to their positions defined in the external reference frame (e.g. camera optical center point) in which the movement of the segments (AR markers) are defined and tracked. This global position is calculated through homogeneous coordinate transformation.

```java
public class Marker
{
    public String name;
    public int segmentID;
    
    public Vector<double> globalCoordinates;
    public Vector<double> localCoordinates;
    
    public bool isVisible;

    public Marker(String name, int segmentID)
    {
        this.name = name;
        this.segmentID = segmentID;
        
        globalCoordinates = Vector<double>.Build.Dense(3);
        localCoordinates = Vector<double>.Build.Dense(3);
        isVisible = false;
    }

    public Marker(Marker markerToCopy)
    {
        this.name = markerToCopy.name;
        this.segmentID = markerToCopy.segmentID;
        
        globalCoordinates = Vector<double>.Build.DenseOfVector(markerToCopy.globalCoordinates);
        localCoordinates = Vector<double>.Build.DenseOfVector(markerToCopy.globalCoordinates);
        isVisible = markerToCopy.isVisible;
    }

    public bool IsCalibrated()
    {
        if (localCoordinates[0] == 0 && localCoordinates[1] == 0 &&
            localCoordinates[2] == 0)
            return false;
        else
            return true;
    }
}
```

**RotMatrices** is a helper class that performs matrix operations with rotation matrices for calculating homogeneous coordinate transformation. In this class the ZH rotatioinal convention is applied with respect to the order of the rotation axes.
During the moment of **calibration of anatomical points** the instantaneous global coordinates of the anatomical point are given by the calibration point of the calibration wand which in this practice acts as a special segment. (In the recording this must be performed when - after palpation - the calibration point is aligned to the calibrated anatomical point of the subject.) The local coordinates of the calibration wand is known as by design, thus the calibration point can always be calculated.
From the instantaneous global coordinates of the anatomical point (given by the calibration point) and the actually tracked coordinates and orientation of the segment the local coordinates of the anatomical point relative to the tracked segment is calculated in the `UpdateLocalCoordinates` function.

```csharp
public bool CalibrateAnatPoint(int markerID)
{
    Marker anatPoint = markerDictionary[markerID];

    if (bodySegmentDictionary.ContainsKey(anatPoint.segmentID))
    {
        Segment parentSegment = bodySegmentDictionary[anatPoint.segmentID];

        if (parentSegment.isVisible)
        {
            Marker calibrationWand = markerDictionary[GaitModel.POINTER_MARKER_ID];

            if (calibrationWand.isVisible)
            {
                UpdateLocalCoordinates(anatPoint, parentSegment, calibrationWand);
                return true;
            }
            else
            {
                throw new System.Exception("The calibration wand is not visible. The anatomical point was not calibrated.");
            }
        }
        else
        {
            throw new System.Exception("The segment for " + anatPoint.name + " is not visible. The anatomical point was not calibrated.");
        }
    }
    else
    {
        return false;
    }
}

private void UpdateLocalCoordinates(Marker anat, Segment parentSegment, Marker calibrationWand)
{
    Vector<double> P_glob = calibrationWand.globalCoordinates - parentSegment.position;
    Matrix<double> R = RotMatrices.invZXZ(parentSegment.orientation);
    Vector<double> P_loc = R * (P_glob);

    anatPoint.localCoordinates = P_loc;
}
```
Calculation of the calibrated anatomical point positions through homogeneous coordinate transformation is continuously performed after the anatomical calibration with the following function.

```csharp
private void UpdateAnatPointCoordinates(Marker AnatPoint, Segment parentSegment)
{
    Matrix<double> R = RotMatrices.ZXZ(parentSegment.orientation);
    Matrix<double> T = DenseMatrix.OfArray(new double[,] {
        {R[0,0], R[0,1], R[0,2], parentSegment.position[0]},
        {R[1,0], R[1,1], R[1,2], parentSegment.position[1]},
        {R[2,0], R[2,1], R[2,2], parentSegment.position[2]},
        {0,0,0,1}});
    Vector<double> P_loc = Vector<double>.Build.Dense(4);
    P_loc[0] = AnatPoint.localCoordinates[0];
    P_loc[1] = AnatPoint.localCoordinates[1];
    P_loc[3] = 1;
    Vector<double> P_glob = T * P_loc;
    AnatPoint.globalCoordinates[0] = P_glob[0];
    AnatPoint.globalCoordinates[1] = P_glob[1];
    AnatPoint.globalCoordinates[2] = P_glob[2];
    AnatPoint.isVisible = true;
}
```