ESTIMATING ANKLE JOINT IMPEDANCE IN THE STANCE PHASE OF GAIT USING AN INSTRUMENTED TREADMILL

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ABSTRACT

Quantifying increased resistance of the ankle joint to motion, referred to as ankle impedance, is important for determining the appropriate treatment for rehabilitation of upper motor neuron (UMN) diseases. The increased ankle impedance can be attributed to increased viscoelasticity of the tissues surrounding the joint clinically treated with splinting and exaggerated stretch reflexes clinically treated with botulinum toxin injections. The choosing of appropriate clinical treatment can be improved if discrimination would be possible between the contribution of exaggerated reflexes and increased tissue viscoelasticity.

The Ashworth Score is a widely used clinical test to quantify the effect of UMN diseases on joint mechanics although it is not able to discriminate between the contribution of exaggerated stretch reflexes and increased tissue viscoelasticity. Moreover, the Ashworth Score is performed under stationary conditions although the clinical treatments are aimed at improving the ability of the patients to perform functional movement such as walking.

Therefore, there is a clinical need to improve the method of quantifying ankle impedance by discriminating between the two contributing factors of exaggerated stretch reflexes and increased tissue viscoelasticity. Specifically, there is a clinical benefit in quantifying ankle impedance in walking to enable the clinical treatment of UMN diseases to be determined on the basis of a functional movement such as walking.

The goal of this study is to design and evaluate a method to identify ankle impedance in the stance phase of gait. Moreover, the contribution of tissue viscoelasticity referred to as intrinsic ankle impedance. An instrumented treadmill that measured forces and moments, a motion capture system that measured positions and a linear model of the moment around the ankle joint was used to estimate the underlying parameters of ankle impedance i.e. damping and stiffness as well as the mass of the foot. To investigate the reliability of the method it was also applied to a 20 kg weight where the mass was the estimated parameter.

For the experiment performed on the weight the estimated mass was 21.7 kg on average with a standard error of the mean (SEM) of 8.3%. For the experiment performed on the ankle joint the mass of the foot was estimated 3.7 kg on average with a SEM of 68%, damping was 2.8 Nms on average with a SEM value of 32.1% and stiffness was 425.1 Nm on average with a SEM value of 28%. The mass of the 20 kg weight could be estimated accurately, while the parameters of the ankle joint were estimated with low accuracy especially the mass of the foot. Overall, the developed method provides a good basis for future work and further measurements.
1 INTRODUCTION

The ankle is a crucial joint in walking, its role in the gait cycle is to adapt to uneven terrain, provide shock absorption during heel strike and transfer a substantial amount of energy at the right instance to propel the body forward [1, 2].

The resistance of the ankle joint to motion is referred to as ankle impedance, which is defined as the dynamic relation between angular displacement of a joint and the resulting torque response [3]. The contributing components to ankle impedance are intrinsic viscoelasticity of the tissue surrounding the joint and the active muscle force including the activity of proprioceptive reflexes. Both the active and passive components depend on the state of the joint, its angle, angular velocity and torque [4-6]. The main difference between the contribution of intrinsic viscoelasticity and the contribution of proprioceptive reflexes to ankle impedance is that reflexes do not change ankle impedance instantaneously because of a time delay in the transport and process time of neural input signals of at least 40 ms for the ankle joint [6, 7].

System identification methods have been widely used to acquire ankle impedance where the change in ankle angle is the input and torque response the output to a mathematical model of the underlying components of inertia, damping and stiffness [3, 4, 8]. The ankle joint is a nonlinear system but in a linear analysis it can be assumed to be linear around a stationary operational point of the system. Perturbations in the form of torque or angular displacement are applied around the operational point to identify the ability of the joint to reject disturbances and therefore quantify the impedance.

In the past, ankle impedance has been identified from steady-state experiments, where the joint is perturbed while the subject is seated. The results have suggested that the ankle joint can be accurately modeled as a second order linear system with underlying components of inertia, stiffness and damping [8]. Moreover, the time delay of reflexes has been used to separate between the two contributions of intrinsic and reflex components to ankle impedance, where the instantaneous response is attributed to the intrinsic component and the delayed response to reflexes [7]. However, the knowledge obtained with seated experiments cannot be directly transferred to walking because ankle impedance is affected by the walking pattern.

Recently, the intrinsic ankle impedance has been identified in walking where the ankle joint is perturbed in the stance phase of gait by rotating the foot with a robotic platform in the walkway [9]. To eliminate the effects of changes in ankle angle and torque around the ankle joint due to walking the mean perturbed angle and torque responses where subtracted from the mean of the trials where the subject was not perturbed [9]. Therefore, only obtaining the responses due to the perturbations.

Impaired walking abilities can be caused by diseases such as stroke and cerebral palsy (CP) where the nerves in the upper part of the central nervous system (CNS) e.g. the brain are damaged. The predominant effect of damaged upper motor neurons (UMN) is muscle weakness (clinically: paresis) and exaggerated stretch reflex (clinically: spasticity). Further, the muscle fibers and the surrounding tissues adapt to these neural changes by increased tissue viscoelasticity [6]. The most widely used clinical method to quantify the effect of UMN diseases on joint mechanics is the Ashworth Scale, where an examiner moves a joint through its range of motion and rates the joint resistance on a scale of 0 to 4 [10]. There are two main approaches for treating UMN diseases. One of them is splinting, which aims at minimizing the effect of increased tissue viscoelasticity the other is botulinum toxin that suppresses the exaggerated stretch reflex. The choice of patient treatment could be influenced if a distinction could be made between the two factors of underlying tissue and reflex properties.

Moreover, the disadvantages of the Ashworth Scale is its subjectivity to the examiner, it is not able to differentiate between the underlying factors of exaggerated stretch reflex and increased tissue viscoelasticity [11] and it is performed under passive conditions although treatments are aimed at improving the ability of the patients to perform functional tasks such as walking. The
benefit of acquiring ankle impedance during walking is that the choice and evaluation of treatment can be directed towards improving the ability of patients to walk. Therefore, there is a clinical need to improve the method of quantifying ankle impedance by discriminating between the two contributing factors of exaggerated stretch reflexes and increased tissue viscoelasticity. Specifically, there is a clinical benefit in quantifying ankle impedance in walking to enable the clinical treatment of UMN diseases to be determined on the basis of a functional movement such as walking.

The goal is to design and evaluate a method to quantify ankle impedance in the stance phase of gait on an instrumented treadmill using translational velocity changes of the treadmill belts to perturb the ankle joint. The torque response was the output and the change in ankle angle due to the perturbations the input to a linear mathematical model of the moment around the ankle joint, from it the underlying parameters of damping and stiffness as well as the mass of the foot were estimated. The linearity assumption restricts the time window to 40 ms after the onset of a perturbation to exclude the effect of reflexes, which introduce the nonlinear behavior of time delays. UMN disease patients have abnormal initial contact to the ground caused by the increased ankle impedance, therefore the perturbations were applied at midstance, when the foot is flat on the ground. To eliminate the effect of the walking pattern on ankle impedance the mean of the unperturbed trials was subtracted from each perturbed trial.

Moreover, the reliability of the method was investigated by estimating the mass of a weight and compare it to its known value of 20 kg.
2 METHODS

2.1 EQUIPMENT

![Diagram of experimental setup]

Figure 1: The experimental setup with the instrumented treadmill and the motion capture system as well as the two host computers one for each system, the position sensor, markers and the xPC real-time computer

2.1.1 INSTRUMENTED TREADMILL

A dual belt instrumented treadmill (ForceLink B.V, Culemborg, The Netherlands) that measures forces and moments was used. Under each belt there were 7 force transducers, 4 measured vertical forces, 2 measured medio-lateral forces (ML) and one that measures anterio-posterior (AP) forces, Figure 2. The transducers are ring torsion load cells with a capacity of 1000 kg (Vishay Revere Transducers, Breda, The Netherlands) and accuracy of +/- 0.02%. The 14 measured signals are multiplied with a 12 x 14 calibration matrix to obtain a total of 12 signals, 3 forces $F_x$, $F_y$, $F_z$ and 3 moments $M_x$, $M_y$, and $M_z$ referenced to the coordinate system of the treadmill.

The treadmill belts are actuated at the rear by two ELAU PacDrive high dynamic SM-140 servo motors (Schneider Electric SA, France) with a capacity of 6.5 kW. The treadmill is controlled in real-time using Matlab/Simulink and xPC target (The Mathworks Inc., Natick, Massachusetts).
Figure 2: On the left a schematic of the treadmill belts showing the location of all force transducers in the coordinate system of the treadmill in mm. The letters inside the circles represent whether the transducer is under the right or the left belt, L and R respectively and X, Y and Z are the directions of measurement. On the right is a figure of the treadmill.

2.1.2 Motion capture system
The motion of the body segments was captured using Optotrak Certus (Northern Digital Inc., Ontario, Canada) motion capture system with a 3D spatial accuracy of +/- 0.1 mm. The position sensor is an optical instrument that detects the infrared light emitted by the marker diodes within the measurement volume. The markers are activated by the strober controlled by the systems control unit (SCU). The marker positions are by default referenced to a global coordinate system located within the position sensor, Figure 1.

2.1.3 Data collection
The treadmill data collection is initialized with a command from Matlab running on the host computer to the xPC, after the experiment the data is sent from the xPC to the host computer and stored. The data from the motion capture system was collected and stored using a software application, First Principles (Northern Digital Inc., Ontario, Canada). The data from the treadmill and the motion capture system were synchronized in time using a trigger signal from the xPC to the SCU.
2.2 EXPERIMENTS

Firstly, the reliability of the forces and moments from the treadmill and the positions of the markers from the motion capture system was investigated by perturbing a weight of 20 kg on the belt and estimating the mass from the measured data. Secondly, intrinsic ankle joint impedance was estimated from the measured data by perturbing the joint at midstance.

2.2.1 PERTURBATION SIGNAL

The perturbation signal, superimposed on the velocity of the treadmill, was chosen with regards to the walking experiment where the goal was to disturb the gait enough to induce a torque response and an angular displacement of the ankle without the subject losing its balance. Therefore, the disturbance should be as small as possible. A sinusoid was chosen since it returns the foot to its initial position after the perturbation. The period of the signal was 0.1 s, which is short compared to the duration of the stance phase, which was approximately 0.7-0.8 in the walking experiment. Further, the smallest amplitude that elicited responses in the torque and ankle angle was chosen after trying several different amplitudes on a walking subject, Figure 3.

![Perturbation signal](image)

**Figure 3:** The sinusoidal perturbation signal that was applied to both the weight and the human

2.2.2 WEIGHT EXPERIMENT

The 20 kg weight on the treadmill was moved with low-frequency sinusoidal velocity of 0.25 Hz with maximum amplitude of 0.3 m/s for 50 seconds. The velocity was determined by visually minimizing any movement of the weight relative to the belt. To resemble a walking experiment, where the velocity of the treadmill was 3 times higher, the perturbation was superimposed at the instance of maximum velocity, Figure 4. The position of the weight was measured by four markers situated on the side facing the position sensor. To keep the weight from slipping on the treadmill belt when a perturbation was applied a double tape was added between the weight and belt.
2.2.3 Walking Experiment

A 29-year-old female subject that weighed 58 kg was instructed to walk normally on the treadmill at a constant velocity of 0.9 m/s during the experimental time of 90 seconds. The subject was barefoot to eliminate any effects of footwear on the measurements. To minimize the effect of soft tissue movement the four markers were placed on bony landmarks of the subject’s right leg and foot facing the position sensor. The landmarks were, lateral malleolus (LM), representing the location of the ankle joint, caput fibulae (CF), the calcaneus (CC) and the proximal fifth metatarsal joint (FM), Figure 5.

Figure 4: On the left is the 20 kg weight on the belt with four markers facing the position sensor. On the right is the commanded treadmill velocity, the perturbation was superimposed at the instance of maximum velocity in the negative y-direction

Figure 5: The figure to the left shows the four bony landmarks with the markers and their corresponding abbreviations. The right figure shows the subject on the treadmill
To determine the timing of the perturbation, the heel strike was first detected from the vertical ground reaction force \( F_z \). To reject noise and accurately detect real time heel strike a discrete moving average filter with a cut off frequency of 20 Hz was applied to the data before verifying the following threshold condition related to the weight of the subject.

\[
\text{th} = 0.05 \cdot m_{\text{body}} \cdot g \quad (1)
\]

Where \( m_{\text{body}} \) is the bodyweight of the subject and \( g \) is the gravitational acceleration. Secondly, to apply the perturbation at midstance a fixed time delay of 0.23 s from heel strike to perturbation onset was chosen by first trying different timings of the perturbation onset and visualizing \( F_z \) in real time.

To further reject noise \( F_z \) was conditioned to be above the threshold for at least 0.002 s before a heel strike was detected. To know when a next perturbation could be applied after the effect of a previous perturbation had vanished the step time right before and after a perturbation was examined. The step time had returned to its previous value after 2.2 steps on average. To eliminate any effects of a previous perturbation at least 4 steps had to pass before the next perturbation could be applied.

### 2.3 Data Analysis

#### 2.3.1 Weight Experiment

Forces were recorded at 1000 Hz sample frequency. Marker positions were sampled at 100 Hz. The raw treadmill and marker data was filtered with a 3rd order Butterworth filter with a cut off frequency of 20 Hz to reduce high frequency noise and the effect of aliasing. After filtering the treadmill data was resampled to 100 Hz to match the data from the motion capture system.

Marker velocities and accelerations were obtained by differentiation of their measured positions. To be able to accurately combine the measurements of force and position the markers were referenced to the coordinate system of the treadmill by defining the treadmill as a rigid body within the measurement space of the markers and defining a new local coordinate system that coincided with the coordinate system of the treadmill.

The application of the perturbation signal affected the force transducers by generating forces within the treadmill that were not part of the response of the weight to the perturbations. Therefore, the inertial forces and moments induced within the treadmill during a perturbation were measured in the unloaded case (with nothing on the belt) and subtracted from the experimental data (see Appendix I).

The force and marker data was segmented into 100-ms-windows of the same time length as the perturbation signal, starting from the onset of a perturbation determined from the y-directional force. For the purpose of only including the effect of the perturbations and not the slow sinusoidal movement of the weight the mean of all the unperturbed windows was subtracted from each perturbed window. Unperturbed windows started at the same instance in the slow sinusoidal movement as if a perturbation had been applied.

The relationship between the force and acceleration in y-direction is assumed to be purely inertial at the initial response phase. Therefore, offline synchronization was performed on the perturbed windows where the first peak in y-directional acceleration of a marker was matched with the first peak in y-directional force (see Appendix II).

A mathematical model of the force on the weight due to the perturbations was derived for the purpose of estimating the mass of the weight. On Figure 6 is a free body diagram of the weight with the relevant forces.
Below, the equations of motion for the weight on the belt are derived. The forces in $y$ and $z$-direction due to the perturbation:

\[ F_z = m_w \cdot (g + \ddot{z}) \]  
\[ F_y = m_w \cdot \ddot{y} \]  

Where $m_w$ is the estimated mass of the weight. The total force becomes:

\[ |\vec{F}_{tot}| = \sqrt{\dot{F}_y^2 + \dot{F}_z^2} \]  

The corresponding acceleration is given by:

\[ |\ddot{a}_{tot}| = \sqrt{\dot{y}^2 + \dot{z}^2} \]  

Finally, the mathematical model of the force on the weight due to the perturbations becomes:

\[ \vec{F}_{tot} = m_w \cdot \ddot{a}_{tot} \]  

Where $m_w$ is the mass of the weight that was the parameter that was estimated by minimizing the difference between the measured $F_{tot}$ and the predicted $F_{tot}$ using the nonlinear least squares function in Matlab.

To eliminate any offset in the data related to the subtraction of the mean unperturbed trace the first value of each window was subtracted from all the data points in the window before estimating the mass of the weight.

### 2.3.2 Walking Experiment

Forces were recorded at 1000 Hz sample frequency. Marker positions were sampled at 300 Hz.

The raw treadmill and marker data was filtered with a 3rd order Butterworth filter with a cut off frequency of 20 Hz to reduce high frequency noise and the effect of aliasing. After filtering the treadmill data was interpolated to 300 Hz to match the data from the motion capture system. Marker velocities and accelerations were obtained by differentiation of the their measured positions.

All marker data was referenced to the coordinate system of the treadmill using the same method as for the weight experiment. Moreover the inertial forces and moments were subtracted from all data as described for the weight experiment.

The timing of heel strike was detected offline using the same threshold as applied in real-time, see Equation 1. The force and marker data was segmented into windows starting from heel strike and ending 0.7 s later, approximately at toe off. For the purpose of eliminating the effect of the
walking pattern and only include the effect of the perturbations in further analysis subtracting the mean of the unperturbed window from the perturbed windows was performed. The data was further segmented into 40-ms-windows starting from the onset of a perturbation, determined from the change in y-directional force. A mathematical model of the moment around the ankle joint was derived for the purpose of estimating the underlying parameters of intrinsic ankle impedance i.e. damping and stiffness. On Figure 8 is a free body diagram of the foot with the relevant forces and moments.

Figure 7: Free body diagram of the foot, shank and the ankle joint in the coordinate system of the treadmill with relevant forces, moments and moment arms. $F_a$ is the force due to the mass of the foot, $F_{grf}$ is the ground reaction force. $M_A$ is the ankle joint moment, $r_a$ is a vector pointing from the origin of the coordinate system to the ankle joint, $r_n$ points to the center of mass of the foot and $r_{grf}$ points to the location of the application of the ground reaction force.

The moment balance equation around the center of mass of the foot:

$$\Sigma \vec{M} : \quad l_f \cdot \vec{\omega} = 0 \quad (7)$$

Where, $\vec{\omega}$ is the angular acceleration of the foot segment and $l_f$ is the moment of inertia about the center of mass of the foot. Summing moments around ankle joint, by applying the parallel axis theorem on the moment of inertia and solving for $M_A$:

$$(l_f + l^2 \cdot m_f) \cdot \vec{\omega} = \left(\vec{F}_{grf} - \vec{r}_a\right) \times \vec{F}_{grf} + M_A + \left(\vec{r}_m - \vec{r}_a\right) \times \vec{F}_m \quad (8)$$

$$\vec{M}_A = -\left(\vec{F}_{grf} - \vec{r}_a\right) \times \vec{F}_{grf} - \left(\vec{r}_m - \vec{r}_a\right) \times \vec{F}_m + \left(l_f + l^2 \cdot m_f\right) \cdot \vec{\omega} \quad (9)$$

Where $l$ is the distance between the ankle joint and the center of mass (COM) of the foot. $M_A$ is equal and opposite of the internal moment, $M_I$ generated in the muscles and the surrounding tissues:

$$\vec{M}_A = -\vec{M}_I \quad (10)$$

$$\vec{M}_I = B\ddot{\theta} + K\ddot{\theta} \quad (11)$$

Where $B$ is damping and $K$ is stiffness of the muscles and the surrounding tissue and $\theta$ is the ankle angle. By introducing $M_I$ the moment balance equation becomes:

$$-B\ddot{\theta} - K\ddot{\theta} = (l_f + l^2 \cdot m_f) \cdot \vec{\omega} - \left(\vec{F}_{grf} - \vec{r}_a\right) \times \vec{F}_{grf} - \left(\vec{r}_m - \vec{r}_a\right) \times \vec{F}_m \quad (12)$$
Rearranging Equation 18 and solving for $M_{grf}$ gives the mathematical model of the moment around the ankle joint, due to the perturbations.

$$
\vec{M}_{grf} = B \vec{\dot{\theta}} + K \vec{\dot{\theta}} + m_f \cdot (I^2 \cdot \vec{\ddot{\omega}} - (\vec{r}_m^e - \vec{r}_a^e) \times \vec{g}) + I_f \cdot \vec{\ddot{\omega}}
$$

(13)

Where, $B$, $K$ and $m_f$ are the unknown parameters to be estimated. $I_f$ was found to be 0.004 kg·m$^2$ from anthropometric data for the subject [12]. The angles and moment arms were calculated from the position of the markers and anthropometric data (see the below subchapters). To eliminate any offset in the data related to the subtraction of the mean unperturbed trace the first value of each 40-ms-window was subtracted from all the data points in the window.

The parameters $B$, $K$ and $m_f$ were estimated from the torque and angular responses induced by the perturbations where $B$ and $K$ represent the intrinsic damping and stiffness of the ankle joint.

The parameters were estimated by minimizing the difference between the measured $M_{grf}$ and the predicted $M_{grf}$ calculated from Equation 13 using the nonlinear least squares function in Matlab. The parameter $m_f$ was estimated as a benchmark that could be compared to anthropometric data and therefore indicate the quality of the model.

**Definition of angles**

The definition of the ankle and the foot segment angles for the mathematical model are shown on Figure 8.

**Figure 8:** The definition of the ankle angle $\theta$ and the foot segment angle $\omega$ from the markers

The ankle angle, $\theta$ was defined between the two vectors pointing from LM to CF (CFLM) and LM to CC (CCLM). The foot segment angle, $\omega$ was defined between the vector CCLM and the z-axis.

Vectors between the markers as well as the z-axis:

$$\vec{r}_{CFLM} = CF - LM$$

(14)

$$\vec{r}_{CCLM} = CC - LM$$

(15)

$$\vec{Z} = \begin{pmatrix} 0 \\ \vdots \\ 1 \end{pmatrix}$$

(16)
The length of the vectors:

\[
|\vec{r}_{CFLM_i}| = \sqrt{\vec{r}_{CFLM_i}^2 + \vec{r}_{CFLM_i}^2} \tag{17}
\]

\[
|\vec{r}_{CCLM_i}| = \sqrt{\vec{r}_{CCLM_i}^2 + \vec{r}_{CCLM_i}^2} \tag{18}
\]

\[
|\vec{Z}_L| = \sqrt{Z_y^2 + Z_z^2} \tag{19}
\]

The segment angle, \(\omega\):

\[
\tilde{\omega} = \cos^{-1}\left(\frac{\vec{r}_{CCLM} \cdot \vec{Z}_L}{|\vec{r}_{CCLM}| \cdot |\vec{Z}_L|}\right) \tag{20}
\]

The ankle angle, \(\theta\):

\[
\tilde{\theta} = \cos^{-1}\left(\frac{\vec{r}_{CFLM} \cdot \vec{r}_{CCLM}}{|\vec{r}_{CFLM}| \cdot |\vec{r}_{CCLM}|}\right) \tag{21}
\]

**Moment arms**

The distance \(l\) from the ankle joint to the COM of the foot was found using anthropometric ratios for the foot segment [12].

\[
r_{COM_y} = \text{Foot length}_y \cdot 0.4485 \quad r_{COM_z} = \text{Foot length}_z \cdot 0.4622 \tag{22 & 23}
\]

Where \(r_{COM_y}\) is the distance from the COM to the ankle joint in y-direction and \(r_{COM_z}\) is the distance from the COM to the ankle joint in z-direction. The lengths of the foot are measured in y-direction from the heel to tip of longest toe and in z-direction from the sphyron to the sole of the foot. The COM locations are from the heel and the sphyron but transferred to the ankle joint to get the correct moment arms from the ankle joint to the COM. The marker locations for the y-direction and measurements from the ankle to the sphyron of the subject for the z-direction were used.

Table 1 shows the values used in Equation 24 to obtain the distance \(l\).

<table>
<thead>
<tr>
<th>COM moment arms</th>
<th>Length [m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>(r_{COM_y})</td>
<td>0.05</td>
</tr>
<tr>
<td>(r_{COM_z})</td>
<td>0.035</td>
</tr>
</tbody>
</table>

\[
l = \sqrt{r_{cm_y}^2 + r_{cm_z}^2} \tag{24}
\]

The movement of the COM in y and z-direction were estimated with by calculating the centroid of the three markers situated on the foot segment.

\[
r_{my} = \frac{(r_{ay} + r_{hy} + r_{cy})}{3} \quad r_{mz} = \frac{(r_{az} + r_{hz} + r_{iz})}{3} \tag{25 & 26}
\]

Where \(r_a\) is the ankle marker, \(r_h\) is the heel bone marker and \(r_i\) is the marker located at the fifth metatarsal.
2.3.3 Quality of Model Fit and Accuracy of Parameters

To quantify how well the two models predicted the measured variables the variance accounted for (VAF) was calculated for both the weight and the walking experiment.

\[
VAF = \left( 1 - \frac{\sum_{i=1}^{n}(y(t_i) - \hat{y}(t_i))^2}{\sum_{i=1}^{n}(y(t_i))^2} \right) \cdot 100\% \tag{33}
\]

Where, \(y(t_i)\) is the measured output and \(\hat{y}(t_i)\) is the estimated output of the model and \(n\) is the number of data points. A high VAF value indicates that the variance in the data can be well represented by the model.

Further, to investigate the accuracy of the fitted parameters the standard error of the mean (SEM) was calculated for all estimated parameters for the weight and the walking experiment.

\[
SEM: \sigma_{\mu_M} = \frac{1}{\sqrt{N}} \cdot P_N \tag{34}
\]

Where \(P_N\) is the parameter covariance and \(N\) is the number of data points in the error function used to estimate the mean. A low SEM value in comparison to the estimated parameter indicates an accurate estimation of its value.
3 RESULTS

The weight and the walking experiments were completed successfully. The results chapter is split up into two sections, one covers the results from the weight experiment and the other the walking experiment.

3.1 WEIGHT EXPERIMENT

A single experiment was analyzed that included 7 perturbed and 7 unperturbed instances.

Responses to the perturbation

Figure 9 shows the measured force and position in y and z-direction. The dotted lines represent the chosen window used for the analysis. The response due to the perturbations was largest in y-direction as expected with a range of 920 N in force and 0.02 m in position. The baseline of the force in z-direction represents the mass of the weight, showing approximately 20 kg, on average the exact weight was expected. The z-directional force response had a range of 20 N. The position response in the z-direction was small but visible.

Figure 9: Forces and positions in y and z directions measured by the treadmill and the motion capture system. The vertical dotted lines represent the chosen window used in the analysis
Figure 10 shows the response to the perturbations in both y and z-direction after segmenting the data into 100-ms-windows and plotting the perturbations on top of each other. The response in the y-direction was larger than for the z-direction as was expected.

Figure 10: Force and acceleration responses due to all the 7 perturbations in both y and z-direction within the 100-ms-window used for further analysis. Top row is force and acceleration in y-direction, bottom row is force and acceleration in z-direction. All the perturbed instances, as well as the mean and standard deviation of the unperturbed traces.
Model fit and parameter accuracy

The VAF value was above 82% for all 7 perturbations, which indicates that the measured force was well described by the model. The mean value of the mass was 21.7 kg (+/- 1.6). The mean SEM value, which represents how accurately the parameter was estimated, was 8.3% of the mean value. All values are summarized in Table 2. Figure 11 shows the fit of the predicted force to the measured force for the 3rd perturbation.

Table 2: Estimated mass, VAF values and SEM values for all 7 perturbations

<table>
<thead>
<tr>
<th>Nr.</th>
<th>Estimated mass [kg]</th>
<th>VAF</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>23.7</td>
<td>99.4</td>
</tr>
<tr>
<td>2</td>
<td>20.4</td>
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<tr>
<td>7</td>
<td>22.0</td>
<td>94.4</td>
</tr>
</tbody>
</table>

Mean 21.7

Std. 1.5

SEM 1.8

Figure 11: The model fit for the 3rd perturbation where the measured force was fitted on a mathematical model of the predicted force.
3.2 WALKING EXPERIMENT

A single experiment was analyzed where there were 10 perturbed and 59 unperturbed steps.

Responses to the perturbation
Figure 12 shows the forces for all the perturbed steps and the mean of the unperturbed steps in the stance phase. Heel strike occurs at time zero and toe off approximately 0.7 seconds later. The vertical dotted lines represent the chosen window of 40 ms, the time before reflexes are able to affect the force response. The chosen window begins at the time of onset of a perturbation. As expected there was a larger response to the perturbation observed in $F_y$ of approximately 30 N and 10 N in the $z$-direction.

In the beginning of stance phase of Figure 12 left graph, the $F_y$ force is negative because the body weight is situated behind the point of contact and the human experiences a negative $y$-directional force from the treadmill as a reaction to its body weight. As time progresses the weight moves over the foot and $F_y$ becomes less negative. At midstance, when the body weight is situated almost directly on top of the foot and $F_y$ is close to zero the perturbation is applied in the negative $y$-direction. The next two peaks, also in $y$-direction, represent the positive forward velocity change and then again negative backwards change in velocity due to the sinusoidal shape of the perturbation signal. In late stance phase $F_y$ is positive, the weight is situated in front of the foot.

On the Figure 12, right graph, the force $F_z$ increases rapidly after heel strike when the foot is being loaded with the body weight. The perturbations at midstance cause a negative initial peak to the backwards velocity change which indicates that the pressure is relieved. $F_z$ shows a later response to the perturbation, the negative peak is visible after 40 ms at the end of the chosen window.

![Figure 12: Forces in y and z-direction, all the perturbed traces, the mean unperturbed and standard deviation for the unperturbed trace. The vertical dotted lines represent the 40 ms window used for the analysis. The heel strike occurs at time zero](image-url)
Figure 13 shows the ankle angle, velocity and acceleration for all the perturbed steps as well as the mean and standard deviation for the unperturbed trace. The ankle angle has a negative peak response that can be attributed to the movement/rotation of the CC marker in the negative y-direction relative to the LM marker causing the ankle angle to decrease. The visible response in the ankle angle within the chosen window is 0.009 rad (0.5°).

Figure 13: Ankle angle, angular velocity and acceleration. All perturbed instances plotted on top of each other as well as the mean and standard deviation of the unperturbed traces. The chosen 40-ms-window used for further analysis is marked with the two vertical dotted lines. Heel strike is at time zero and toe off approximately after 0.7 seconds.
The change in $M_{grf}$ due to the perturbations is shown on Figure 14. On the left graph the stance phase is shown where the heel strike occurs at time zero and toe off approximately after 0.7 s. On the right graph is a closer look at the chosen window after subtracting the mean unperturbed trace from each perturbed trace. The range of change in the $M_{grf}$ due to the perturbations was approximately 15-20 N.

Figure 14: The moment around the ankle joint due to ground reaction forces, all perturbed instances and the mean unperturbed trace with the standard deviation. On the left graph heel strike occurs at time zero. On the right graph there is a closer look at the timing of a perturbation after subtracting the mean unperturbed trace. The dotted vertical lines represent the perturbation window

Model fit & parameter accuracy
The VAF value was above 84% for all 10 perturbations. The mean values for the estimated parameters, was 3.7 kg (+/- 3.5) for the mass, 2.8 Nms/rad (+/- 1.5) for damping and 425.1 Nm/rad (+/- 143.2) for the stiffness. The SEM value for the mass was 68.0% of the mean value, 32.1% for damping and for stiffness 28.0%. All values are summarized in Table 3. Figure 15 shows the fit of the predicted moment around the ankle joint due to the ground reaction force to its measured value for the 8th perturbation.

Table 3: The estimated parameters for mass of the foot and damping and stiffness of the ankle joint for all 10 perturbations with mean, standard deviation (std.), VAF and SEM values.

<table>
<thead>
<tr>
<th>Nr.</th>
<th>$m_f$ [kg]</th>
<th>$B$ [Nms/rad]</th>
<th>$K$ [Nm/rad]</th>
<th>VAF</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.2</td>
<td>2.3</td>
<td>533.9</td>
<td>97.6</td>
</tr>
<tr>
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<td>7.4</td>
<td>5.0</td>
<td>465.9</td>
<td>94.9</td>
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<td>3</td>
<td>4.2</td>
<td>1.6</td>
<td>495.5</td>
<td>94.2</td>
</tr>
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<td>10.5</td>
<td>1.3</td>
<td>83.0</td>
<td>95.6</td>
</tr>
<tr>
<td>5</td>
<td>0.2</td>
<td>5.4</td>
<td>338.5</td>
<td>84.4</td>
</tr>
<tr>
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<td>2.2</td>
<td>2.4</td>
<td>395.1</td>
<td>93.9</td>
</tr>
<tr>
<td>7</td>
<td>-1.3</td>
<td>1.3</td>
<td>548.0</td>
<td>99.8</td>
</tr>
<tr>
<td>8</td>
<td>4.9</td>
<td>2.0</td>
<td>570.2</td>
<td>99.2</td>
</tr>
<tr>
<td>9</td>
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<td>2.1</td>
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<td>10</td>
<td>2.2</td>
<td>4.1</td>
<td>379.8</td>
<td>88.2</td>
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<tr>
<td>Mean</td>
<td>3.7</td>
<td>2.8</td>
<td>425.1</td>
<td></td>
</tr>
<tr>
<td>Std.</td>
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<tr>
<td>SEM</td>
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<td>0.9</td>
<td>118.7</td>
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Figure 15: Model fit, for the 8th perturbation where the predicted moment due to ground reaction forces was fitted to a model of the measured moment due to ground reaction forces.
4 DISCUSSION

The aim of the study was to design and evaluate a method of using an instrumented treadmill and a motion capture system to estimate intrinsic ankle impedance in the stance phase of gait. For this purpose a sinusoidal perturbation was superimposed on the velocity signal of the treadmill. A mathematical model of the moment around the ankle joint was derived and the underlying components of damping and stiffness were estimated as well as the mass of the foot. Moreover, the reliability of the method was investigated by applying it to a 20 kg weight.

Reliability of the method
The VAF value for the weight experiment of above 82% for all the perturbations shows that the variance in the data was fairly well represented by the model. The low SEM value of 8.3% of the mean estimated mass indicates that the optimization procedure was able to estimate the parameter well.

The VAF value for the walking experiment of above 84% for all the perturbations shows that the model could account for a large part of the variance in the data where 8 out of 10 VAF values were above 90%. The estimated mass of the foot was a measure of the methodological reliability of the walking experiment. The mean value of 3.7 kg (+/- 3.5) was over 4 times higher than the value of 0.86 kg from anthropometric data [12]. The overestimation of the mass as well as the negative value of the 7th perturbation indicates what the SEM value of 67% reveals, that the optimization procedure was not able to accurately estimate the parameter from the measured data. There is a possibility that the mass was moving in a way not accounted for by the model such as slip on the belt. Moreover, the fast perturbations could have caused deformation in the foot especially during the first instances after onset of a perturbation causing the change in the ankle angle to be smaller than if the foot was a rigid body.

The parameters of damping and stiffness had lower SEM values, therefore the optimization procedure was able to estimate their values more accurately then the mass of the foot although they were high compared to the weight experiment.

The filter cut off frequency of 20 Hz was rather low for an experiment including a perturbation signal of 10 Hz. The effect of choosing such a low cut off frequency is that some of the measured signal is thrown out. However, the force data was noisy and to accommodate the data analysis where the onset of a perturbation was detected from the force in y-direction the low cut off frequency was needed.

Effect of natural variance in gait
A possible source of error in the walking experiment causing lower VAF values is the method of subtracting the mean of the unperturbed traces from each perturbed trace. Studies on the spatial and temporal parameters of gait in normal walking have shown that the variance of normal gait is not random, which is the assumption made by taking the mean [13]. Consecutive steps have been found to be inter-related in terms of timing of peaks [14], therefore the subtraction of the last unperturbed step before a perturbed one might be more suitable.

Comparison to literature
In a recent comparable study the parameters of intrinsic damping and stiffness were estimated at a fixed time of 0.225 s following heel strike, the mean value was 0.58 Nms/rad for damping and 145 Nm/rad for stiffness for a subject of the same weight as used in the current study [9]. The perturbation signal was in the form of 2° dorsiflexive ramp that lasted for 75 ms and induced a torque response of approximately 2 Nm. The values for damping and stiffness are higher in the current study a mean of 2.8 Nms/rad for damping and 425.1 Nm/rad as well as the angular displacement was lower, approximately 0.5° and the torque response in the range of 15-20 Nm. It has been found in a seated experiment that stiffness and damping of the ankle joint decreases
with increased displacement amplitude [15] which can explain the different values obtained in the two studies.

Limitations
The current method utilizes a linear model of the moment around the ankle joint to acquire the intrinsic ankle impedance and therefore leaves out the contribution of reflexes. The time delay in transporting neural signal causes the need for a nonlinear model to estimate the contribution of reflexes.

Future recommendations
Future research should expand the method of estimating intrinsic ankle impedance to include the contribution of reflexes and therefore enable discrimination between the intrinsic and reflex components.
In the future higher filter cut off frequency should be used to make sure that no signal is lost, that could be done by e.g. measuring the actual velocity of the treadmill and use it to estimate the perturbation onset.
The accuracy of the method can be further investigated by applying the perturbation at e.g. later instances in time after heel strike and compare the parameters of damping, stiffness and mass of the foot. Ankle impedance is expected to increase from early stance to late stance [9] due to increased muscle activation needed for propelling the body forward. However, the mass of the foot should remain the same.
5 CONCLUSION

The method of using an instrumented dual belt treadmill, a motion capture system and a linear model of the moment around the ankle joint to obtain the intrinsic ankle impedance in the stance phase of gait was investigated.

The method was successful in estimating the mass of a 20 kg weight, which was used as a measure of its reliability. The mass of the foot could not be accurately estimated, stiffness and damping was estimated with slightly higher accuracy. Possible reason for the low accuracy of the estimated parameters for the walking experiment is the assumption that the foot is a rigid body under the experimental conditions and the natural variance of gait. Overall the method provides a good basis for future improvements and further measurements.
6 REFERENCES

APPENDIX I – TREADMILL DYNAMICS

The application of a perturbation can affect the force transducers by generating forces and moments within the treadmill that are not part of the response of the weight or the human walking on the treadmill. Therefore these effects need to be subtracted from the experimental data.

The forces and moments from the treadmill were measured in the unloaded case at a constant velocity of 0.9 m/s, the same velocity as for the walking experiment. The perturbation signal was superimposed on the constant velocity every 2 seconds. The sampling frequency was 1000 Hz. The raw data was filtered with a 3rd order Butterworth filter with a cut off frequency of 20 Hz. The treadmill data was resampled to either 100 or 300 Hz to match the data from each of the experiments.

All data was segmented into 100-ms-windows starting with the onset of the perturbation determined by the commanded velocity of the treadmill.

In total 5 perturbations were measured, the mean for each force and moment component within the perturbation window is shown on Figure 16. The forces and moments were subtracted from each perturbation window analyzed in both the weight and the walking experiment.

Figure 16: Upper graph shows the forces and lower graph the moments induced within the window of 100 ms that are subtracted from the experimental data
APPENDIX II − OFFLINE SYNCHRONIZATION

The initial data analysis for the weight experiment revealed shifts in time between the peaks of acceleration and force in y-direction. Figure 17 shows the force and acceleration data where perturbation onset is at time zero and the data has been segmented and each perturbation plotted on top of each other. The shifts observed in the acceleration traces on Figure 17 do not coincide with the properties of the weight and were therefore eliminated before fitting the data to the model.

The source of the problem was found in the hardware of the SCU for the motion capture system. The SCU of the motion capture system and the real-time computer (xPC) of the treadmill have separate internal clocks. The internal clocks are responsible for giving each data point a time stamp. One second in the motion capture system was shorter than one second in the real-time computer therefore the data from the motion capture system moves ahead of the data measured by the treadmill. This causes the acceleration peaks to move ahead of the force peaks and the data to fit badly to the model as time progresses.

A firmware update was provided by the producers of the motion capture system, which reduced the shifts. The offline synchronization was only performed on the data from the weight experiment since the data of the walking experiment was collected after the firmware update. Table 4 shows the estimated mass and the VAF values before and after offline synchronization for the 7 perturbations as well as how many samples each perturbation was shifted.

![Figure 17: Acceleration and force in y-direction after cutting the data](image)

Table 4: The table shows the 7 instances of perturbation for the first and last value of the mass experiment as well as the last value before and after the data was synchronized offline.

<table>
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APPENDIX III – CENTER OF PRESSURE CALCULATION

Weight experiment
As a secondary investigation of data from the treadmill and the motion capture system the center of pressure (COP) was calculated from the treadmill data and compared to the centroid of the 4 markers on the weight representing the approximate location of the (COM). The two measurements are different but their movement on the belt should coincide for the rigid weight when there is no movement in the z-direction. The COP was calculated using the forces and moments from the treadmill using the following equations.

\[
COP_x = -\frac{M_y}{F_z} \quad (A1) \\
COP_y = \frac{M_x}{F_z} \quad (A2)
\]

The centroid of the four markers was calculated as an estimate of the COM from the motion capture system.

\[
COM_x = \frac{(m_{1x} + m_{2x} + m_{3x} + m_{4x})}{4} \quad \text{COM}_y = \frac{(m_{1y} + m_{2y} + m_{3y} + m_{4y})}{4} \quad (A3 & A4)
\]

Where \( m \) represents each of the 4 markers and the subscript (1 – 4) and the measured direction (\( y \) or \( z \)) respectively.

The location of the COP in the \( y \)-direction calculated from the treadmill forces and moments and the estimated movement of the COM between the four markers on the weight were similar for the slow sinusoidal movement as shown on the left graph of Figure 18. On the right graph is a closer look of the instance of perturbation, the motion capture system measures 0.02 m excursion and the treadmill a 0.24 m. The excursion of the COP is close to the length of the weight in \( y \)-direction of 0.23 m.

![Figure 18](image)

**Figure 18:** On the left graph the location of the COP in \( y \)-direction during the slow sinusoidal movement. On the right graph a closer look at the time of a perturbation

Figure 19 shows the movement of the COP in \( x \)-direction for both the slow sinusoidal movement and a closer look at the timing of a perturbation. There are two main differences between the results. Firstly there is an offset of 0.05 m between the two systems because the markers are located on the side of the weight closer to the position sensor than the COP that is approximately under the center of the weight.

![Figure 19](image)
Secondly, the motion capture system is showing a slow sinusoidal movement that might be misalignment between the coordinate system of the markers and the treadmill. Thirdly, a response to the perturbation in the x-direction is not visible from the motion capture system but the treadmill data shows an excursion of 0.08 m most likely due to the acceleration of the weight and/or small tilting effects.

Figure 19: Both graphs show the movement of the COP in x-direction from the treadmill data and the motion capture system. The dotted line represents the time of onset of a perturbation. The right figure shows a closer look of the response due to the perturbations

Walking experiment

The COP is located at the application of the $F_{gfrf}$. The y-directional location is of interest since it should be located within the foot of the subject and can therefore be used as a secondary investigation into the accuracy of the method. The moment arm from the ground reaction force to the ankle joint is:

$$r_{gfrf} = COP_y - LM_y \quad (A5)$$

On Figure 20 the moment arm of the ground reaction force is shown, the perturbed traces show that the excursion of the ground reaction force is around 0.05-0.07 m which is within the y-directional length of the foot of the subject, which was measured to be approximately 0.25 m.

Figure 20: The moment arm of the ground reaction force
APPENDIX IV – SENSITIVITY ANALSYS

To better understand how each parameter contributes to the model fit one parameter at a time was varied in magnitude from 10% of its optimized estimated value up to 100% in intervals of 10% while the other parameters were kept constant and plotted with the predicted $M_{grf}$.

Figure 21. The stiffness is the most effective parameter where the fit to the model shows the largest discrepancy from the predicted $M_{grf}$ when it is only 10% of its optimized value.

Figure 21: Sensitivity analysis for the 8th perturbation, each parameter is varied from 10-100% from its optimized value while the other parameters are kept constant.
**APPENDIX V – CALIBRATION MATRIX**

The following 12x14 calibration matrix that was used in the data analysis to obtain the 3 forces, $F_x$, $F_y$, $F_z$ and 3 moments $M_x$, $M_y$, $M_z$ for each belt.

\[
\begin{pmatrix}
-27.948 & -19.73 & 4.3379 & 8.1379 & -981.23 & -982.4 & -5.5999 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
6.6373 & -19.859 & -1.9313 & -12.028 & -31.003 & 39.462 & -1123.6 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
558.88 & 498.16 & 501.77 & 468.15 & -33.585 & 8.0569 & 21.453 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
-374.29 & 364.11 & 316.24 & -375.21 & 7.2096 & 12.672 & -165.84 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
306.7 & 267.07 & 14.231 & -10.247 & 19.028 & 42.389 & 19.777 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
26.374 & -12.319 & 14.801 & 0.85588 & -810.54 & 770.23 & 756.8 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & -16.53 & 34.849 & -3.0132 & 115.53 & 1019.7 & 984.23 & -21.969 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & -8.7102 & 41.836 & 22.625 & 53.46 & 57.912 & -69.837 & -1066.3 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & 468.17 & 506.57 & 501.94 & 543.1 & -5.4265 & -2.8001 & -21.211 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & -305.5 & 371.02 & 374.5 & -341.4 & -17.053 & -15.597 & -165.61 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & 3.0936 & -21.761 & -264.94 & -317.87 & -26.344 & -23.91 & 21.81 \\
0 & 0 & 0 & 0 & 0 & 0 & 0 & 21.035 & -36.526 & 3.568 & 33.642 & -856.76 & 873.57 & -745.85
\]