MASTER THESIS

Human lower limb gait analysis of walking on uneven terrain

Jesse Kaiser
s1355783

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Biomedical Engineering
Department of Biomechanical Engineering

<u>Exam Committee</u>
Dr. E.F.H. van Asseldonk
W.F. Rampeltshammer, Msc.
Dr. Ir. G.A. Folkertsma

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Abstract

Nowadays exoskeletons are primarily used in controlled environments. In order to increase the applicability of exoskeletons by using them outside of these controlled environments, they need to have a robust controller capable of handling outside terrain and obstacles. Therefore, this research aims to find lower limb gait parameters able to detect uneven terrain without a delay to be implemented in an adaptive exoskeleton controller. Additionally, research was performed on finding kinematic and kinetic input for desired lower limb joint angle and moment trajectory generation. To accomplish this, experiments were performed with ten healthy subjects while stepping on a beam. Kinematics and kinetics were measured using inertial measurement units and a dual force plate situated under the beam. The experimental conditions consisted of three different foot placements (forefoot, midfoot and rearfoot), three different beam heights (32mm, 58mm and 81mm) and two beam orientations. Results showed that the leading ankle, knee and hip angle could be used as parameters to detect beam height and foot placement before or at heel strike, while the center of mass height could detect these during mid stance. Furthermore, kinetic results showed a delayed moment onset for the rearfoot condition in the leading knee. The forefoot condition showed an increased joint moment compared to level, with the largest increase in moment amplitude for the leading ankle joint. Using a linear discrimination method proved that early beam detection is possible using kinematic data. The gathered kinematic and kinetic data could also provide input parameters for desired trajectory determination, usable in exoskeleton control. This information can be the basis for human inspired adaptive controller for lower limb exoskeletons.

1. Introduction

Exoskeletons assist humans with physical activities, like walking or lifting objects. Lower limb exoskeletons can be used for multiple applications including gait rehabilitation [1], balance improvement [2–4], load carrying [5] and metabolic cost reduction [2,6,7]. Available exoskeletons are mostly used in a controlled environment, where normal walking and handling obstacles are pre-programmed and 'detection' is performed by the user manually choosing another gait mode [8–10]. Using a controller that can detect and react to different obstacles autonomously, could improve the usability of exoskeletons in general. This would enable using exoskeletons in unknown environments outside of the lab during daily life activities. For example, these daily life situations can consist of ascending or descending a slope, stair walking, standing still or walking over uneven terrain.

The importance of these tasks is also reflected in the fact that they are a part of the Cybathlon, a competition where physically disabled people use technological assistance systems to complete everyday tasks. This research is focused on the uneven terrain obstacle, which consists of half round beams of different heights and placement. As previously mentioned, these obstacles are now primarily pre-programmed [8–10]. Creating an adaptive controller would result in these exoskeletons to be more versatile, consequently this means that besides generating a desired torque or angle trajectory using input parameters, the controller must detect the different obstacles to react accordingly. Detection of different gait modes could be possible by using kinematic data [11].

Literature on current control methods was investigated to indicate if and how this would be possible. Jiménez-Fabián et al. concluded in their review paper [12] on the control of ankle orthoses, prosthesis and exoskeletons, that high-level control predominantly consists of using a finite-state machine. They state that this results from the fact that the gait pattern is predominantly cyclic and can be divided into sub-phases, each for which a different mechanical behaviour of the ankle can be distinguished. Kinetic and kinematic information has been used to detect these different phases, mainly with the use of pressure sensors and angle sensors. They also indicated that correctly identifying the transition between different gait phases and different gait modes remains a problem, resulting in the gait modes still being changed manually [8–10]. Li et al. [11] investigated automatic gait mode detection for stair and ramp walking with an ankle-foot orthoses. They used vertical foot position to distinguish between level and stairs and foot pitch to distinguish between level and ramp walking. Detection was performed using optimized thresholds. A severe limitation was the one step detection delay of the controller, which could seriously affect stability. Automatic gait mode detection without delay could increase the usability of exoskeletons, thereby becoming more intuitive and widely applicable, especially in unknown terrain.

Still, for this to be achieved further information on the human biomechanics of walking on uneven terrain is needed. These insights can be useful in the development of new and improvement of existing controllers used in daily life situations. Different kinematic studies have been performed on humans traversing uneven terrain, although the definition of the term 'uneven terrain' is broad and consists of: Beams of different heights [13], tiles of different heights [14–17], loose rocks, inclinations [18] and sand and grass [19–25]. Tiles of random heights showed minor adaptations in stepping strategy, however an increased muscle activity, increased knee and hip joint work [14] and an increased ankle angle variability were observed [14, 17]. When walking on a loose rock surface, subjects lowered their center of mass (CoM) and enlarged their base of support (BoS) to increase stability [19, 26]. On slippery surfaces, subjects adopted their gait to keep their CoM centred over the supporting limb and increased their limb stiffness [20–22]. This indicates the wide variety of uneven terrain types and even larger variety in corresponding adaptations. Showing that not a single strategy can be applied for each type of uneven terrain, addressing the need for further research.

Panizzolo et al. [13] did study changes in gait strategies while stepping on a beam of 48mm and 32mm in height with different foot placements. It was demonstrated that an increased ankle moment is used when landing on a beam with the forefoot and an increased knee and hip moment is used when landing on the beam with the rearfoot. Unfortunately, no specific parameter was found indicating walking over a beam useful for beam detection. Neither did they perform measurements on the trailing leg, midfoot placement and the steps prior to the beam hit, thereby possible excluding useful information.

Current research lacks information on automatic detection of uneven terrain. The previously addressed points also emphasize the need for sufficient kinetic and kinematic information to improve control methods for exoskeletons, including automatic gait mode adjustment for different obstacles. Literature presented studies on different types of uneven terrain, although only limited on walking over a beam. Therefore the aim of this research is to study the detection and gait patterns of beam walking by measuring and analyzing lower limb kinematics and kinetics. In this way the following main research questions can be answered: Which lower limb gait parameters can be used for the detection of stepping on a beam and which lower limb kinematic and kinetic information can be used as input in the control of an exoskeleton walking over a beam? Thereby distinguishing between forefoot, midfoot and rearfoot placements and different beam heights. Kinematics will be measured of both legs and kinetic data of the leading leg. This information could indicate how the human body responds to these kind of perturbations, which can be the basis for human inspired control strategies for exoskeletons.

The structure of this paper is as follows: Section 2 describes the experimental setup, protocol and data analysis. Section 3 presents the kinematic and kinetic results and additionally the outcome of a real-time beam detection test. These results are discussed in Section 4 and the conclusion of this research is presented in Section 5. Extended information on the results are shown in the Appendix.

2. Materials and methods

2.1. Subjects

Two female and eight male subjects (height 185.4±9.8 cm; weight 78.2±9.6 kg; age 23.3±2.2 years; mean±SD) participated in this study. All participants were healthy and had no musculoskeletal injuries or other musculoskeletal diseases and provided informed consent prior to the study. The study was approved by the EEMCS Ethics committee of the University of Twente (Enschede, Netherlands).

2.2. Experimental setup

During this experiment it was investigated how human biomechanics are affected by stepping on a half round beam. To determine viable beam shapes and heights for the experiment, preliminary tests were performed. The results showed that for any beam higher than 81mm it would not be possible to distinguish foot placement, since the beam was too high for the forefoot and rearfoot condition. Using a square beam did not show a significant difference to behaviour on a half round beam, but only provides a larger contact area, increasing the stability during the midfoot condition. To also study the effect of this instability, half round beams were chosen to resemble the uneven terrain. It resembles stepping on tree roots and the uneven terrain used in the Cybathlon 2020 also consists of half round beams in the same height ranges and in a rotated configuration. Midfoot conditions and rotated beam conditions were not tested in experiments in literature, but are still relevant for uneven terrain walking and are therefore included in this research.

Concluding from the preliminary study, wooden beams of three different heights were used: 32mm, 58mm and 81mm with a radius of 33mm. The beams were attached with Lycra strips to a dual force plate (AMTI Force and Motion, MA, USA), which measured ground reaction forces and moments with a sample frequency of 1kHz. The force plate was integrated into a walkway (5.70m x 0.6m) to create an even floor height. The walkway consisted of a wooden frame filled with hard foam plates. Full body kinematics were measured using inertial measurement units (IMUs) of the Xsens Link system (Xsens, Enschede), with a sampling frequency of 240Hz. A total of 17 IMU were used. They were placed on the foot, the shin bone, lateral side above the knee, the sacrum, sternum, scapula, lateral side above the elbow, lateral side of the wrist, back of the hand and backside of the head, according to the Xsens user manual.

2.3. Experimental protocol

Participants started with their right heel in the center of the force plate and were then asked to walk up and down the walkway for three minutes at their preferred walking speed. They were instructed to look at a point at the end of the room while walking, to decrease focus on the beam and stimulate a normal walking situation. The subjects were instructed to step on the beam with their right foot as illustrated in Fig. 3. This procedure was repeated for 14 different conditions: level walking (no beam); Three different beams (32mm, 58mm and 81mm) each with three different foot placements (forefoot, midfoot and rearfoot); And two beams (32mm and 58mm) while rotated 30 degrees with respect to the walking direction with two different foot placements (forefoot and rearfoot) as shown in Fig. 1 and 2. Before each trial the subject had time to determine the best starting positions at each end of the walkway. The conditions were randomized for each subject and before each trial the participants were informed on which foot placement would be used. Wrong foot placements were eliminated during data processing.

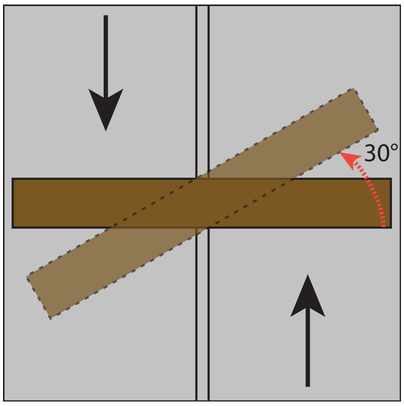

Figure 1: Top view of the force plate with both beam placements. The arrows indicate the walking direction. The red arrow indicate the direction of rotation.

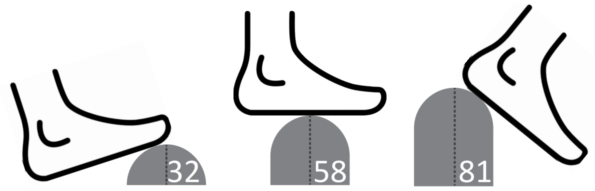

Figure 2: Representative figure of the different foot placements and beam heights from left to right: forefoot (32mm), midfoot (58mm) and rearfoot (81mm).

2.4. Joint kinematics & kinetics

Kinematic data was measured using inertial measurements units from the Xsens Link system. A full body setup was used for each measurement. Full body kinematics were measured during the experiment including position, velocity and acceleration of lower limb body segments and joints. Lower limb joint angles were appropriately segmented for two gait cycles using Matlab. One before and one after beam hit. Heel strikes were detected using the acceleration peak of the right foot. In the processing phase a beam hit was detected by using the heel strikes and vertical center of mass position. Next to the joint angles and accelerations, the Xsens software provided virtual marker positions. These virtual markers are important in the process of estimating kinetic data. Kinetics were determined using the inverse dynamics tool in OpenSim [27]. OpenSim is software for analyzing and simulating the neuromusculoskeletal system. The human model used in OpenSim was scaled to the body length, foot length and weight of the subject. The unscaled model is 3-dimensional, consists of 23 degrees-of-freedom featuring lower limb and lower back joints and includes 54 muscles.

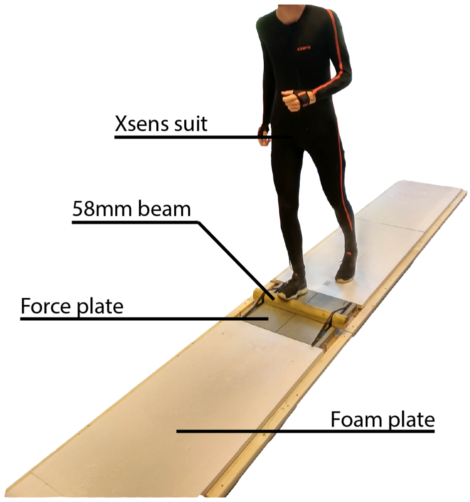

Figure 3: Experimental set-up where a subject, while wearing the Xsens suit, is stepping on the 32mm beam with the forefoot condition. The beam is fastened to the force plate with lycra.

To synchronize the two measurement systems, the force plate and Xsens system were connected using the Xsens syncing station. When starting the Xsens measurement, a pulse was send to Labview NXG, simultaneously starting the force plate measurement. After the measurements, the sampling frequency of the force plate was downsampled to 240Hz.

Obtaining kinetic data with inverse dynamics by combining data from the IMUs and the force plate presented several difficulties. These difficulties arise from the alignment of the coordinate frame of the Xsens with the force plate coordinate frame, described as the global coordinate frame. Firstly, the position data from the IMUs showed rotational drift around the vertical axis in the Xsens coordinate frame. Secondly, starting position errors resulted in a lateral and sagittal offset between the origins of the Xsens and global coordinate frame. Lastly, the in-

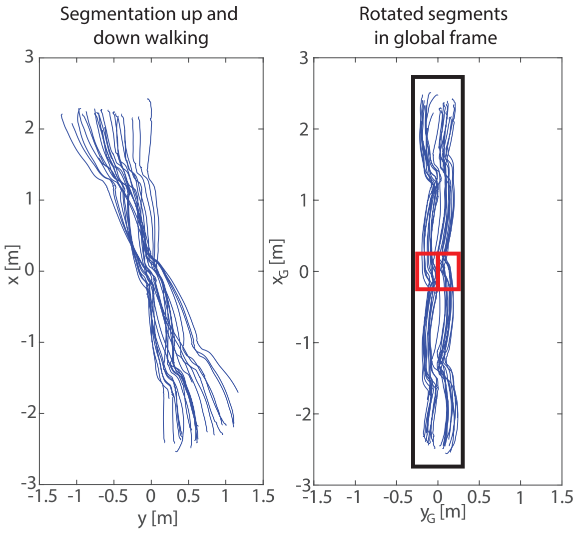

Figure 4: On the left: Right foot position segmentation of walking up and down the walkway, showing rotational drift around the z-axis. Sagittal movement is represented by 'x' and lateral movement by 'y'. On the right: Right foot position from rotated segments of walking up and down with respect to the walkway. The back lines indicate the borders of the walkway, the red boxes indicate the double force plate. Sagittal movement in the global frame is represented by 'x_G', lateral movement in the global frame by 'y_G'.

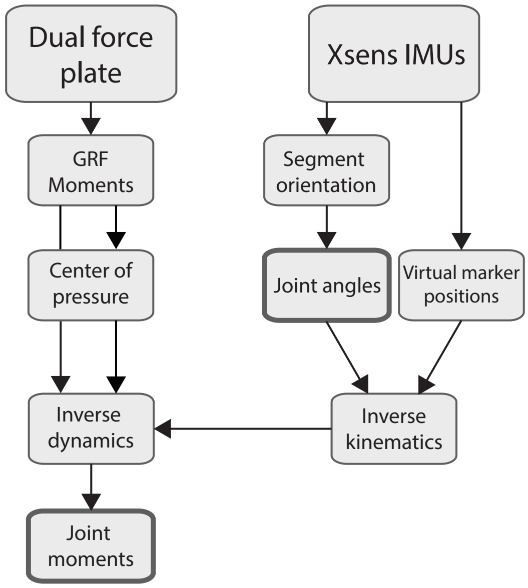

Figure 5: Flow chart of the data analysis. Kinematic data presented in the results section is generated from the IMU measurements. Kinematic data used for the inverse dynamics calculation is determined using virtual marker data and the joint angles from the IMUs. The bold boxes indicate the output results that are analyzed and discussed in this research. GRF, ground reaction forces.

verse dynamics tool was only able to perform inverse dynamics on the first few segments of each trial, the subsequent segments showed an error. This most likely happened due to drift of the virtual marker data, since the error stated that the values in the sagittal plane must be strictly increasing.

The rotational drift can partly be solved by performing a 'good' calibration of the IMUs. The quality of the calibration is stated by the Xsens software, where a good calibration shows the forward and backwards movement along the x-axis. After a good calibration drift does still occur, as shown on the left of Fig. 4. The remaining drift was removed by first segmenting the trials of walking up or down using the right foot position. Next a linear fit was drawn through these segments, after which they were rotated, such that the movement in the sagittal plane aligned with the x-axis of the Xsens coordinate frame. Now that the drift was resolved, it was important to get the ground reaction point forces on the correct CoP under the foot. To best align the origins of the two frames, the subjects were instructed to start each trial with their right heel on the center of the force plate. Unfortunately, this was not always correct, resulting in an offset of the two origins. This was solved by first determining the CoP in the force plate frame using the measured GRFs and moments. By combining the motion of the subject from Xsens data and the CoP of the force plate in OpenSim, the offset could be visualized. The lateral offset was reduced by setting the lateral CoP movement from the virtual heel marker to the virtual toe marker. The sagittal offset was manually reduced by sight, until at heel strike the CoP was position under the heel.

A flow chart of the data analysis is shown in Fig. 5. To perform inverse dynamics, the following inputs are needed: center of pressure, ground reaction forces, moments on the force plate and motion data. The CoP data was derived as previously described.

Ground reaction forces and moments were measured by the force plate. Since only using joint angles as input for motion data is not sufficient, the motion data was generated by using the inverse kinematics tool in OpenSim. The motion data contains time histories of the generalized coordinates that describe the movement of the model. Input for this tool was the virtual marker data, the plantar and dorsiflexion of the ankle and flexion and extension of the knee and hip of both legs. The joint angles were included to improve the beam walking movements, especially movement of the ankle joint. All of the inputs previously stated were then gathered for each segment of up or down walking separately and used in OpenSim, to produce the lower limb joint moments.

2.5. Real-time beam detection

The observed kinematic differences indicated variables for the detection of a beam around heel strike. Indicating that a beam could be detected before or just at the point of hitting a beam, making it possible for controllers to adapt to a perturbation with minimal delay. To show the effectiveness of these parameters, they were tested for each subject individually using a simple threshold method. For each subject 50% of the data was used to determine subject specific threshold values for different foot placements and beam heights. The detection process proceeded as follows. Firstly, a beam hit was detected. After which the detected beam hits were further distinguished between forefoot, midfoot and rearfoot conditions. Lastly, for each foot placement separately different beam heights were distinguished. The parameters used for threshold determination follow from the kinematic results.

2.6. Step length and step time

Step length and step time were determined using the right foot position data from Xsens. The steps before hitting the beam and after hitting the beam were segmented. From the segments, the average values and standard deviations were calculated for each trial. Then the step length and step time for all trials were averaged over all subjects. This information was used to indicate if step adjustments occur due to traversing the beam.

2.7. Statistical analysis

Ankle, knee and hip angles of both legs were statistically tested using a multiple t-test at each percentage of gait to assess differences between the beam conditions and level walking. Also statistical testing was performed to find differences in lower limb joint angles in between foot placements and beam heights. No correction for multiple comparisons was taken into account. This test was chosen to indicate gait cycle areas where beam conditions differed from normal walking. Differences around beam hit could be interesting for beam detection.

Statistical testing on the step length and time was also analyzed using multiple t-tests comparing each beam condition against level walking to find differences related to step adjustments. Correction for multiple comparisons was taken into account. Significance level was set at $P \leq 0.05$ for all analyses.

3. Results

3.1. Kinematics

Fig. 6 & 7 present the kinematic differences between stepping on a beam and level walking. A negative gait phase indicates the step before hitting the beam, where beam hit is defined at 0%. Right ankle angle showed significant differences for all trials compared to level walking around beam hit. For the forefoot and rearfoot condition the range of significance is dependent on the height of the beam. A higher beam showed a bigger range of significant difference. When comparing different foot placements, there was a significant difference between all foot placement conditions. Between forefoot and rearfoot and between midfoot and rearfoot this occurred before heel strike till mid swing. For beam heights there was a significant difference between all beams for the rearfoot and forefoot condition cleary presented in Fig. 7. Only the forefoot condition showed significant difference around beam hit.

Significant differences for the right knee compared to level were found for all beam conditions. Significance ranged from before beam hit to heel off and depended on beam height. However, no significant differences were presented between beam heights and foot placements. The right hip angle showed significant differences for the 58mm beam with the rear- and midfoot conditions compared to level and for the 81mm beam for all foot placements, indicating the effect of beam height on the hip angle. The rearfoot showed significance before beam hit. A significant difference was also found between beam heights for the midfoot and rearfoot condition from before beam hit to mid stance.

Results of the leg stepping over the beam (trailing leg) showed that the left ankle and left knee had significant differences for the 58mm and 81mm trials compared to level walking from mid stance to toe off. The left hip did not show any noticeable (less than 5% of gait) significant differences compared to level walking. The center of mass height presented significant difference for the midfoot condition from beam hit to pre-swing for all beam heights compared to level and the rearfoot condition showed significant differences from before beam hit to toe off for all beam heights illustrated in Fig. 8. A significant difference was found for the rearfoot condition between the 81mm and 32mm beam from foot flat to pre-swing.

30 degrees beam rotation showed no significant differences compared to the same beam heights and foot placements without beam rotation. Indicating that rotation does not influence the plantar- and dorsiflexion of the ankle joint and the flexion and extension of the knee and hip joints.

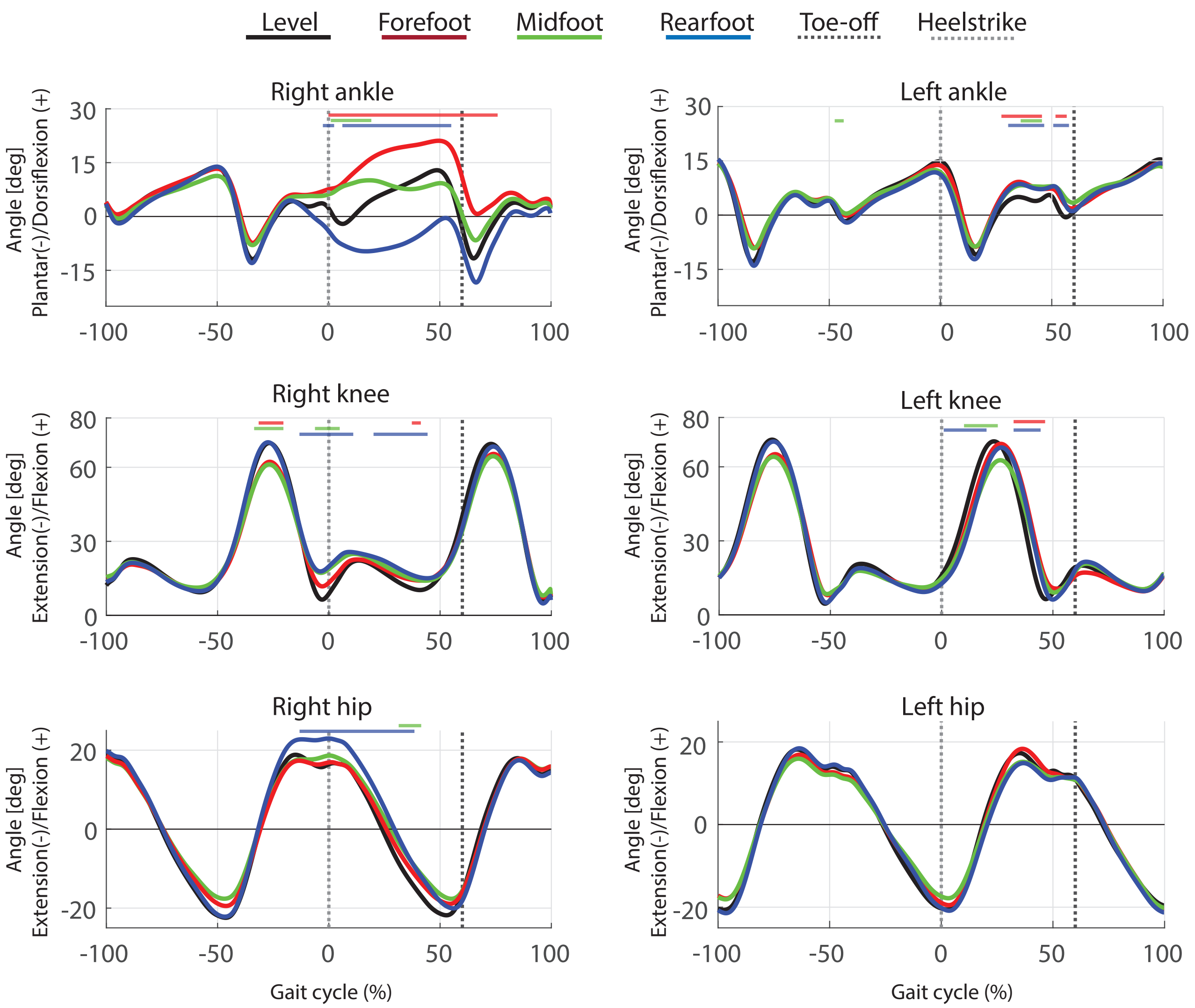

Figure 6: Results of the lower limb joint kinematics. The plots show the mean across beam heights (including the rotated conditions) and subjects for each foot placement. Positive angles represent flexion (dorsiflexion at the ankle) and negative angles represent extension (plantarflexion at the ankle). Representation of 200% gait, where the beam is hit at 0%. Negative value means the cycle before and positive after the beam hit. The vertical dotted lines indicate heel strike (0%) and toe-off (60%). The bars at the top of the figures indicate statistical significance ($P \leq 0.05$) of the 58mm beam compared to level, with the colours responding to the foot placement.

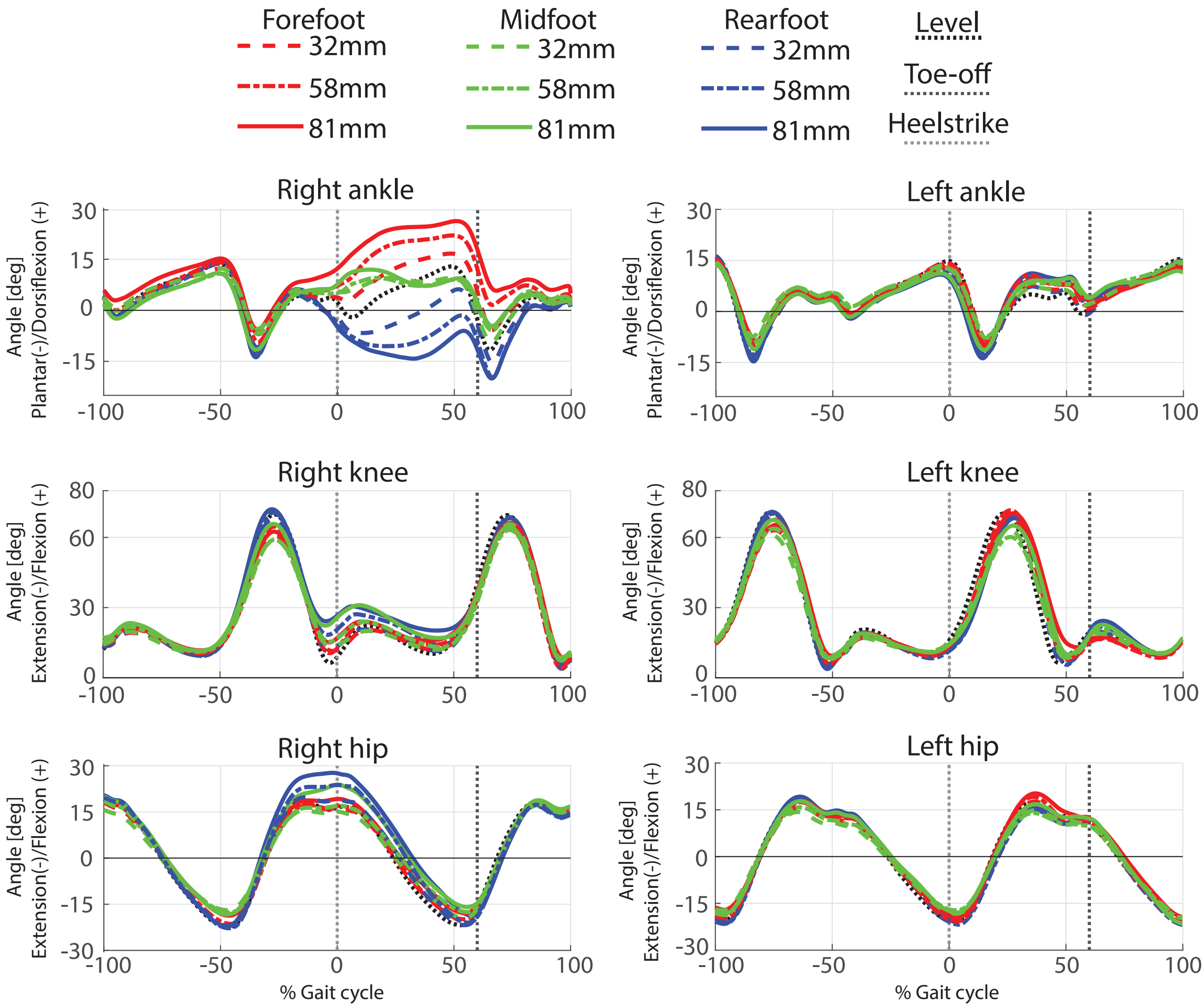

Figure 7: Results of the lower limb joint kinematics. The plots show the results for each beam height and for each foot placement. Positive angles represent flexion (dorsiflexion at the ankle) and negative angles represent extension (plantarflexion at the ankle). Representation of 200% gait, where the beam is hit at 0%. Negative value means the cycle before and positive after the beam hit. The vertical dotted lines indicate heel strike (0%) and toe-off (60%).

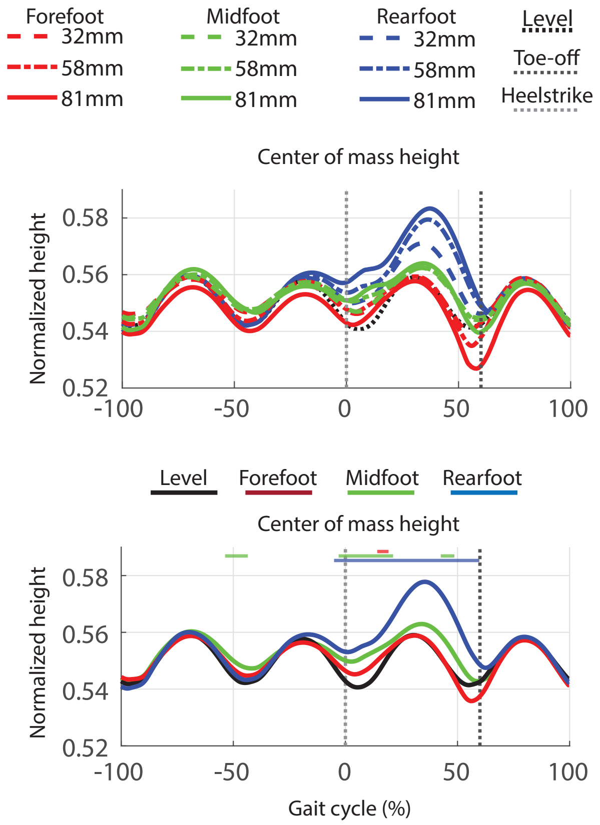

Figure 8: Center of mass height. Top figure shows the results for each beam height and foot placement. The bottom figure shows the mean of all beam heights for each foot placement. The height is normalized to body height. Representation of 200 % gait, where the beam is hit at 0%. The vertical dotted lines indicate heelstrike (0%) and toe-off (60%). The bars at the top of the figures indicate statistical significance (P $\leq$ 0.05) of the 58mm beam compared to level, with the colours responding to the foot placement.

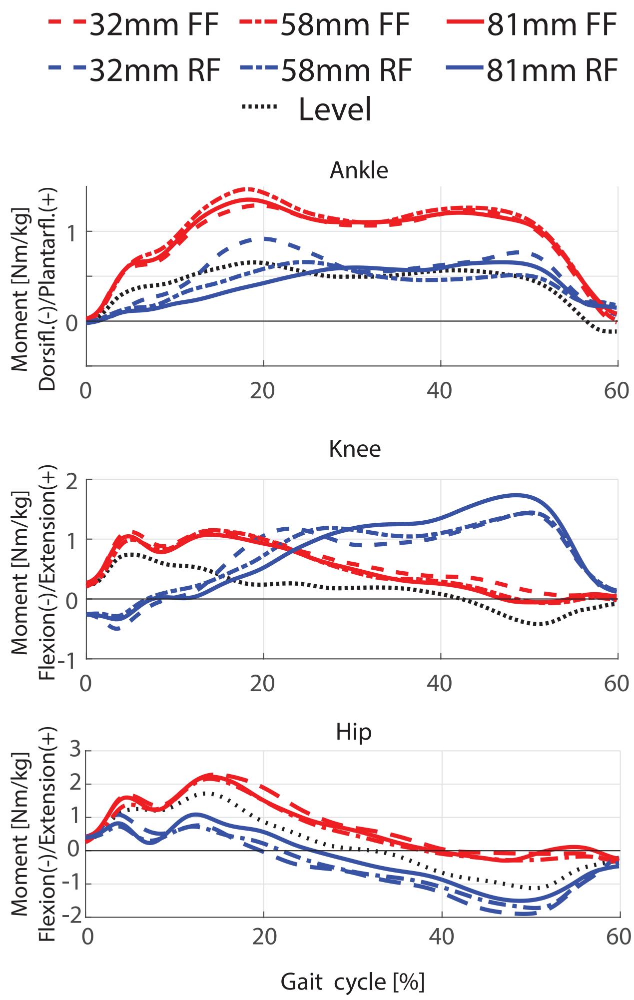

Figure 9: The lower limb joint kinetics of one subject during stance phase (0% to 60% of gait) of the right leg for the three different beam heights and the forefoot and rearfoot conditions. Positive moments indicate extension (plantarflexion at the ankle) and negative moments indicate flexion (dorsiflexion at the ankle)

3.2. Kinetics

The results of the joint moments have been processed for one subject and are presented in Fig. 9. Since the original results of the rearfoot condition for the ankle and knee did not seem logical, they were inverted. Further elaboration on this is provided in the discussion. Due to the low number of usable segments, no statistical testing was performed. Standard deviations of the measured kinetics are presented in Appendix C to indicate variability between measured segments.

The results show that for each joint the forefoot condition resembles the level condition with an increased moment amplitude. During the forefoot condition the ankle first exerts a plantarflexion moment up until pre-swing and the knee and hip joint exert an extension moment until midstance. For the ankle and hip the rearfoot condition torque trajectory resembles that of level walking and the knee exerts a delayed extension moment peak.

3.3. Real-time beam detection

The results of the real-time beam detection are presented using confusion matrices, to both illustrate correct and incorrect beam detection. This could provide additional insight into the effect of this method. The kinematic results presented areas of significance around heel strike from level walking, between beam heights and between foot placements. These differences indicate the possibility for detection between level and beam walking. From these results the following gait areas were used for the different thresholds. Parameters used for the pure beam detection were the right ankle and knee angle from -4% to 0% (heel strike) of gait.

The ankle angle was used for forefoot and rearfoot detection and beam height detection during the forefoot condition, CoM height for midfoot detection and hip angle for beam height detection during the midfoot and rearfoot conditions. The following areas of gait cycle were taken for threshold determination:

Foot placement detection

Predicted [%] \ Actual	FF	MF	RF	Level
FF	71	55	9	12
MF	4	12	17	0
RF	8	20	69	7
Level	17	13	5	81
Correct	71%	12%	69%	81%
Total	580	300	662	138

Beam height detection

Predicted [%] \ Actual		FF 32	FF 58	FF 81	MF 32	MF 58	MF 81	RF 32	RF 58	RF 81
FF	32	42	4	11	18	7	4	5	6	9
FF	58	26	68	31	32	24	35	1	4	3
FF	81	6	19	33	23	21	26	0	1	1
MF	32	7	3	3	13	9	0	20	2	1
MF	58	2	0	0	2	12	2	4	8	2
MF	81	0	0	0	0	0	7	0	3	12
RF	32	3	1	14	8	6	4	13	0	1
RF	58	4	1	0	2	9	0	37	15	1
RF	81	10	4	8	3	12	22	20	60	70
Correct		42%	68%	33%	13%	12%	7%	13%	15%	70%
Total		158	172	36	62	85	115	208	236	186

Figure 10: Upper: Confusion matrix of the foot placement detection. Lower: Confusion matrix of the beam height detection. The columns show the actual foot placement, beam height or level condition, and the rows refer to the condition that was predicted using the thresholds values. The diagonal shows correct detection in percentages. Lower rows show percentage of correct detection and total amount of actual trials. FF, forefoot; MF, midfoot; RF, rearfoot.

Ankle angle (-2% to 2%); ankle angle beam height detection (2% to 5%); hip angle (-6% to 0%) and CoM height (9% to 14%). After the threshold values were determined, they were tested on the other half of the subjects data.

Foot placement detection and beam height detection are presented in Fig. 10. Using this simple linear method correct detection for forefoot, rearfoot and level condition was between 70% and 80%. The midfoot condition showed a correct detection of 12%, which is lower than a correct foot placement detection due to chance. After foot placement detection, beam heights were detected according to foot placement. The most accurate beam height detection was shown for the forefoot condition between 33% and 68%. The midfoot condition only showed a correct detection of around 10%, which equals a beam height detection due to chance. Fig. 10 also indicates that most of the midfoot conditions were detected as a forefoot condition. The rearfoot condition showed a large number of 58mm beam segments detected as 81mm beam segments.

3.4. Step length and step time

Statistical testing of the beam trials compared to level walking showed a significant difference for several condition, although no specific trend was observed. Appendix D shows the results of the step length and time for each trial averaged over all subjects and indicate the specific trials showing statistical significance.

4. Discussion

The aim of this study was to find parameters for the detection of uneven terrain and input parameters for the control of an exoskeleton walking over uneven terrain, specifically beams. Additionally, finding parameters that could indicate different foot placements and different beam heights. To find these, experiments were performed measuring kinematic and kinetic data of the lower limb joints. The results showed significant differences for the joint angles of the leading leg before and around beam hit between the beam conditions and level walking, indicating the possibility for early beam detecting using kinematic data.

4.1. Kinematics and Kinetics

The kinematic results present the largest trajectory deviations from the level condition for the leading ankle. These results are in agreement with the study of Panizzolo et al. [13]. Unfortunately, Panizzolo et al. presented no data from before impact with a beam. Next to the deviations from level walking, significantly increased dorsiflexion was observed between beam heights during forefoot condition. Increased dorsiflexion was also observed by Earhart et al. [18] for an increasing wedge inclination, which is comparable to an increasing beam height. In this research the most important findings are the significant differences around impact with the beam. Fig. 6 shows an average value over all subjects and clearly indicates these differences, although as can be seen in Appendix B trajectories in between subjects can highly differ. Still, with the notice of the average values, a real-time beam detection was performed using the same areas of the gait cycle to determine threshold values for each subject separately. The ankle was used to detect a beam, distinguish between forefoot and rearfoot and between beam heights during the forefoot condition. The results presented in Fig. 10 indicate that, although ankle angle trajectories deviate between subjects, using the same detection area provides promising results for early beam detection. Also the areas of significance provide information on the areas of gait where adjustments to the angle trajectory need to be made during control. Trajectory adaptations are especially required during the rearfoot and forefoot condition for the subject to cross such an obstacle naturally. The midfoot condition shows large deviations between subjects and as indicated in Fig. 10 is mostly mistaken for the forefoot condition. Therefore, the need to account for the midfoot condition will be subject specific.

Next to the ankle, the knee and hip also provide useful information for early detection. The knee shows significantly increased flexion before beam hit. Therefore, the knee was also used to distinguish between beam and level walking. The hip angle shows a significant increased flexion for the rearfoot condition, compared to level, as well as significant differences between beam heights for the mid and rearfoot condition. This is explained by the increasing height of the lower leg, when standing with the heel on the beam and occurs before impact with the beam to increase the ground clearance for the leading leg to step over the beam. This information is useful for early beam detection, therefore the hip angle was used to detect between different beam heights for the mid and rearfoot condition. Fig. 10 shows that correct beam height detection for the midfoot condition is low and most likely occurred due to chance. For the rearfoot condition the highest beam could be detected accurately. The lower beams were not detected, this could be due to wrong threshold value determination. Still, these results show that the hip angle could be used for beam height detection, although improvements to the detection method are required.

The CoM height showed a significant increase in height for the rearfoot and midfoot condition compared to level. This is caused by the foot or heel being placed on the beam. As shown in Fig. 8 the difference between midfoot and level is rather small, but was still used to detect the midfoot condition. This resulted in a low correct detection rate of 12%, which probably occurred due to chance. Next to this, a lot of midfoot conditions were detected as forefoot. This is not surprising when comparing the angle trajectories in Appendix B, which clearly show that the ankle angle for the midfoot and forefoot are similar around beam hit for various subjects. These results indicate that another parameter should be used to detect the midfoot condition, which could be the angle of the foot with respect to the ground.

The significantly increased ankle dorsiflexion and increased knee extension of the trailing leg, indicate that the subjects increased their ground clearance to get the trailing leg over the beam. From this insight, control could be adjusted such that after hitting a beam, regardless of the foot placement, the ground clearance of the leading leg should be increased. As presented in Fig. 7 this is independent of beam height.

Rotating the beams showed no significant differences compared to the straight beams for the same heights and foot placements. Explained by the fact that the beams are of the same height, it is expected that the ankle, knee and hip angle in the sagittal plane are similar. Differences in the eversion and inversion angle of the ankle were found between the rotated and straight beams, this could be relevant for control if exoskeletons were able to actively assist in these directions.

Kinetic results in Fig. 9 are presented for the leading leg after hitting the beam. Deviating torque trajectories from level walking indicate the need for trajectory adaption when hitting a beam. Variating beam height did not demonstrate significant effects on the joint moments, indicating that this information is not necessary for torque trajectory control for these beam heights.

The original kinetic results of the ankle and knee during the rearfoot condition have been inverted, since they showed illogical results and it was expected that during the inverse dynamics calculation of the rearfoot condition the sign was changed. This could be due to the way OpenSim performs their inverse dynamics calculations. Since the equation of motion is dependent on the joint angles, a negative joint angle could result in a sign change of the joint moments. The joint angles are indicated with q in Eq. 1 with M the system mass matrix, C the Coriolus en centrifugal forces and G the gravitational forces.

$$\boldsymbol{M}(q)\ddot{q} + \boldsymbol{C}(q, \dot{q}) + \boldsymbol{G}(q) = \tau \tag{1}$$

In a normal standing position the ankle angle is in OpenSim defined as zero degrees, dorsiflexion is positive and plantarflexion negative. During level walking and the forefoot condition, the ankle angle during stance phase is always positive as shown in Fig. 6. But during the rearfoot condition, the ankle angle during stance is always negative. This could have resulted in an inverted ankle moment trajectory during the rearfoot condition and could also have influenced the inverted knee moment trajectory. Unfortunately, no information on the precise calculation of the inverse dynamics in OpenSim was found to substantiate this claim.

The kinetic results of Panizollo et al. [13] show a plantarflexion moment of the ankle, which increases until pre-swing for all conditions. In this research the ankle shows an increasing plantarflexion until midstance. In the research of Panizollo a mean over multiple subjects is given, the difference could therefore be explained by the subject specific gait pattern in this research. The knee presents similar trajectories for the level and forefoot condition, the rearfoot condition shows a delayed extension peak moment. Panizollo et al. did also report significant later moment onset in the knee, but around 30% of gait, where in this research this occurs around 50%. This could also be explained by the subject specific gait strategy of the subject in this research. The hip shows a similar trajectory for all conditions, with a decreased moment during rearfoot condition. Resulting from the fact that the leg is already over the beam and the ankle is in push-off position.

Comparison to kinetic data of stair walking [28, 29] presents a resemblance in the moment trajectories, especially in the knee where the forefoot condition is comparable to ascending a stair and the rearfoot condition to descending. Riener et al. [28] note that during stair walking the subject makes contact with the forefoot in contrast to level walking, where there is heel contact. This explains why stair ascending resembles the forefoot condition and stair descending the rearfoot condition, because it results in similar ankle joint angles. Stair walking does show increased angle deviations for the hip and knee joint compared to walking over a beam [28, 29]. This indicates that uneven terrain walking shows similarities in kinematics and kinetics of stair walking, which can help with the design of exoskeleton control for different obstacles.

Data that could also improve the use of an exoskeleton for both legs, are the kinetics of the trailing leg. Further study on the trailing would be of interest since it is expected that, especially in the forefoot condition, the push-off power of the trailing ankle increases in order to get the CoM over the beam.

All of these insights could help with generating desired trajectories for the control of an exoskeleton. The largest deviations have been presented for the ankle, which could have a large impact on the experience for the user. The knee and hip of the leading leg showed smaller, although significant differences. For trajectory generation these should be taken into account, but could have less effect on the user. Foot placement and beam height is not important for the trailing leg and the main focus there should be on increasing the ground clearance.

Before generating the desired angle trajectories, correct foot placement and beam heights should be detected. Correct foot placements are specifically important for the leading ankle and hip and for correct torque trajectory generation. Correct beam height detection is specifically important during the forefoot and rearfoot condition for the leading ankle angle and during the rearfoot condition for the leading hip angle, since here the largest angle deviations from level walking occur. The torque trajectories showed no significance for different beam heights for the leading leg, although beam height could affect forces in the leading leg since lifting the CoM would require more force. But Fig. 8 only shows differences between beam height for the rearfoot condition. This could be explained by the beams not being sufficiently high enough for a significant torque increase to occur. These results provide insight to focus on correct foot placement detection, rather than beam heights specified in this paper. However, larger beam heights could induce larger differences.

4.2. Kinematics & kinetics using IMUs

Kinematic information from IMU measurements showed resemblance to research using optical tracking systems [13, 14]. Two studies that performed comparison of an IMU system against optical tracking found root mean square errors of four to three degrees over all joints [30, 31], indicating that an IMU system is a viable option for kinematic measurements. However, determining kinetics using the

Xsens system was more challenging. These challenges included: Rotational drift in the IMU position data, offset between the origins of the Xsens coordinate frame and global coordinate frame and an error in the inverse dynamics tool.

Due to the use of IMUs a rotational drift occurred during each trial, depending on calibration quality. This raised difficulties with the alignment of the Xsens coordinate frame with the coordinate frame of the force plate. This was to most extent solved by splitting the trial into segments and rotating these according to the "real" walking directions, meaning along the x-axis as illustrated in Fig. 4. The error between the origins of the two coordinate frames could be reduced by finding the exact origin of the force plate and start each measurement with the right heel on this point. To find the exact origin, a point force could be exerted on the force plate until the measured moments are zero. During these experiments subject were told to put their heel against the beam in the center of the force plate, but since the beam was also in the middle, there was always a sagittal offset. Using a double force plate made correcting for a starting position error more complicated, since the error could be different for each plate. Also due to the beam the forces were distributed over both plates, sometimes resulting in a small force on the plate not stepped on. The forces of both plates were not combined, since the exact origin of the two plates was unknown. Using a single force plate could reduce the effort to find the right starting position and account for any errors of the CoP estimation, since this would result in one point force and not two. Of course, a lateral offset could still occur in both cases. Center of pressure determination could also be improved by placing an IMU on the force plate, such that the Xsens and force plate coordinate frame alignment could be improved and have a reference to each other. During kinetic data processing only the first few segments were used, since the subsequent segments showed an error. The virtual marker placements are based on an anthropometric database and the body segment lengths of the subject. Since movement of these segments are measured with IMUs, drift in the virtual marker data could occur accounting for the error.

4.3. Step length & step time

For some conditions a significant difference was found for the step length and time compared to level, although not showing any specific trend. This shows that step adjustments occur, but that they are non specific for any beam height and/or foot placement. The small area after the beam where subjects can walk can be a possible explanation. Subjects probably slowed down after the beam hit, affecting the results. Also habituation to a starting position resulting in a specific foot placement, could have reduced step adaptations.

5. Conclusion

The joint angle measurements presented significant differences between beam and level walking. both for different foot placements and beam heights. Therefore, it can be concluded that the ankle, knee and hip joint angles and CoM height can be used as parameters for the detection of uneven terrain before or at heel strike, resulting in minimal controller delay. A beam detection test using these kinematic parameters showed promising results and the possibility for early beam detection. The use of a simple linear method validated the use of specific detection parameters, bringing the design of an adaptive controller a step closer.

Kinetic results showed that foot placement does have an influence on the lower limb joint torque trajectories, meaning that foot placement is important information when developing a controller. Different beam heights did not show a significant effect on the torque trajectories. Concluding that the torque trajectories during different foot placements could be used as input parameters for an exoskeleton controller to determine desired torque trajectories. The use of IMUs for kinetic measurements is a new method and showed difficulties, especially with CoP determination. Although results were obtained for one subject and thus further research needs to be performed.

Limitations of this research consist of not measuring the kinetics of the trailing leg, which could show important information when stepping on a beam. Also no kinetic measurements were performed for the midfoot condition. Consequently, for a complete overview of the human biomechanics while walking over uneven terrain these parameters should be further investigated.

In conclusion, the information gathered with this research improves the understanding of human walking over uneven terrain and could be the basis for the development of human inspired adaptive control methods.

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Appendices

A. Standard deviation of the lower limb joint angles and CoM height

To present variability in the measured kinematic data the following figure is provided. This shows the averaged standard deviation over all beam heights and subjects for each foot placement. Largest deviations are shown during beam hits in the ankle angle.

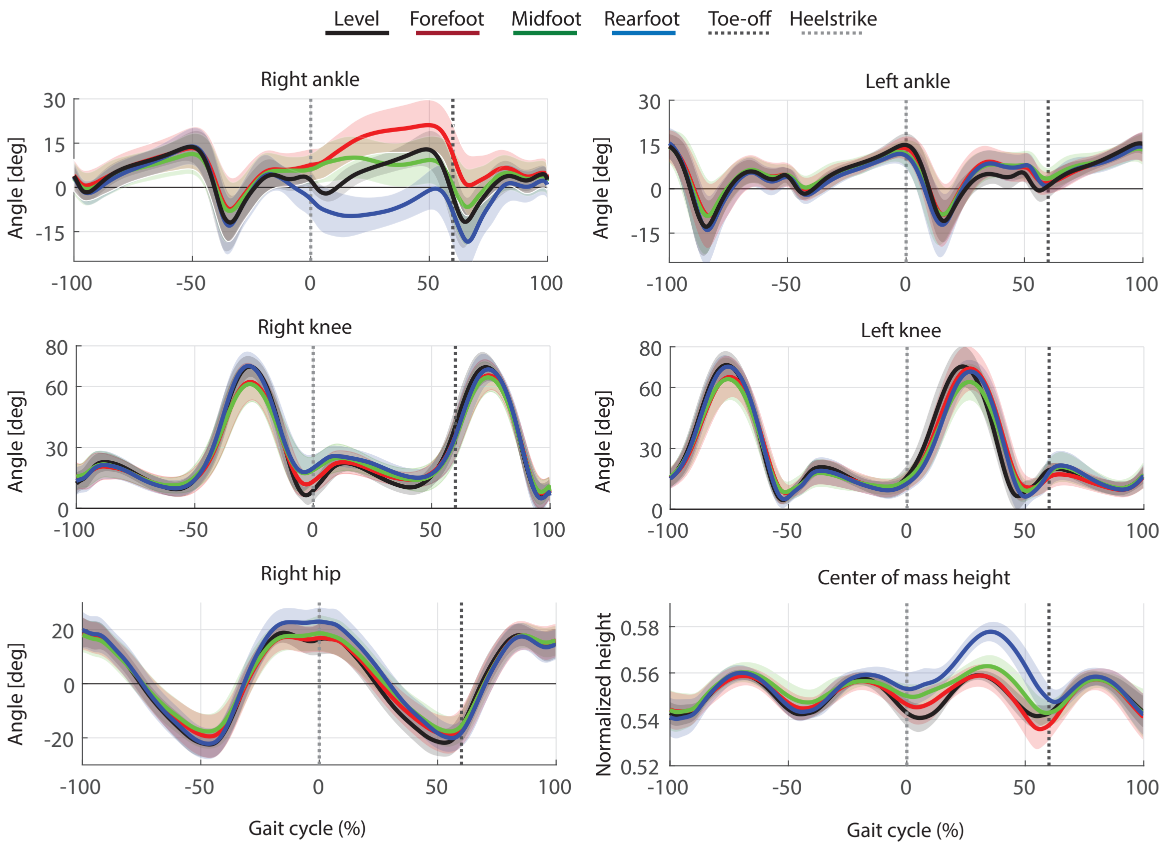

Figure 11: Results of the lower limb joint kinematics and the center of mass height. The plots show the mean across beam heights (including the rotated conditions) and subjects for each foot placement. The shaded areas show the averaged standard deviation of all beam heights for each foot placement. Positive angles represent flexion (dorsiflexion at the ankle) and negative angles represent extension (plantarflexion at the ankle). Representation of 200% gait, where the beam is hit at 0%. The vertical dotted lines indicate heelstrike (0%) and toe-off (60%).

B. Ankle angle of each subject

To show variability between subjects, the ankle angle of the leading leg is shown for each subject. Most noticeably is the midfoot condition which can differ greatly between subjects and sometimes partly resembles the forefoot condition.

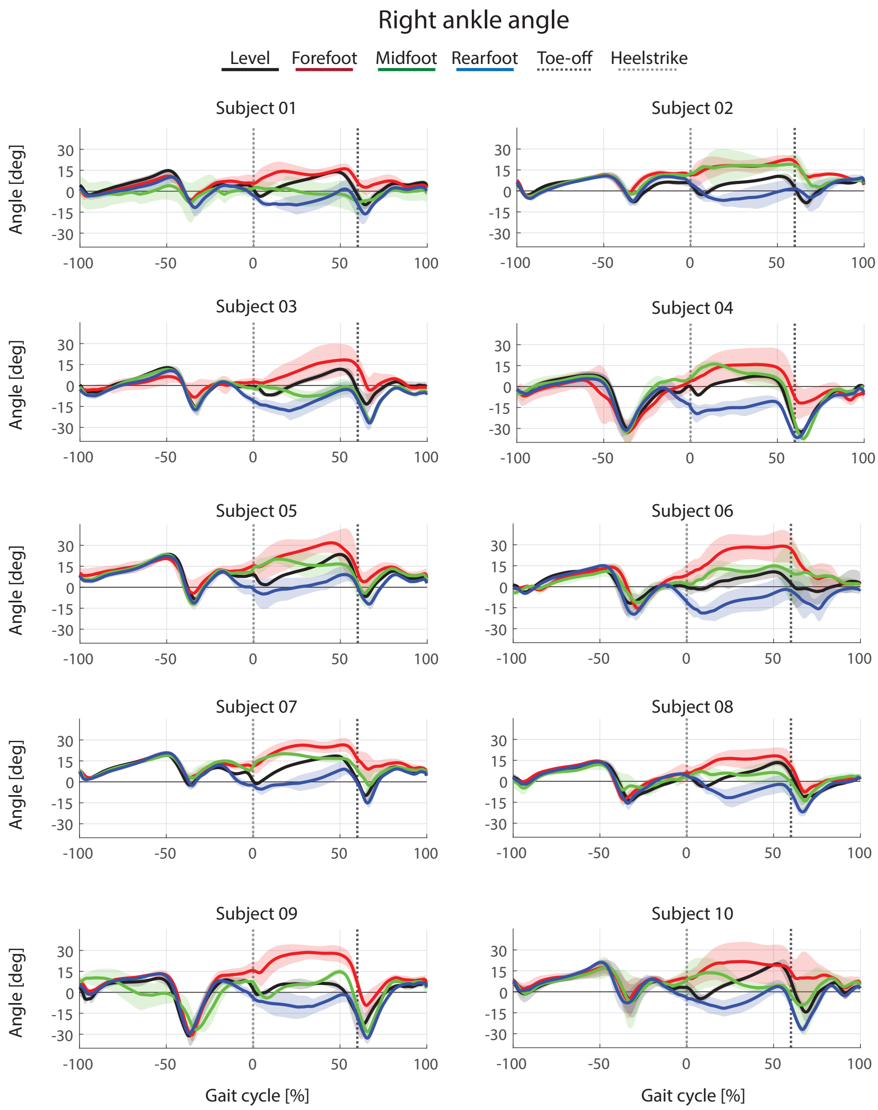

Figure 12: The right ankle angle of each subject to show variability in the data. The plots shows the mean over all beam heights for each foot placement. The shaded areas indicate the SD between the different beam heights. Positive angles represent dorsiflexion and negative angles represent plantarflexion. Representation of 200% gait, where the beam is hit at 0%. The vertical dotted lines indicate heelstrike (0%) and toe-off (60%).

C. Standard deviation of the joint moments

Since kinetic measurements by combining IMUs and a force plate is a novel method, standard deviations of the results are presented to show variability of the measurements.

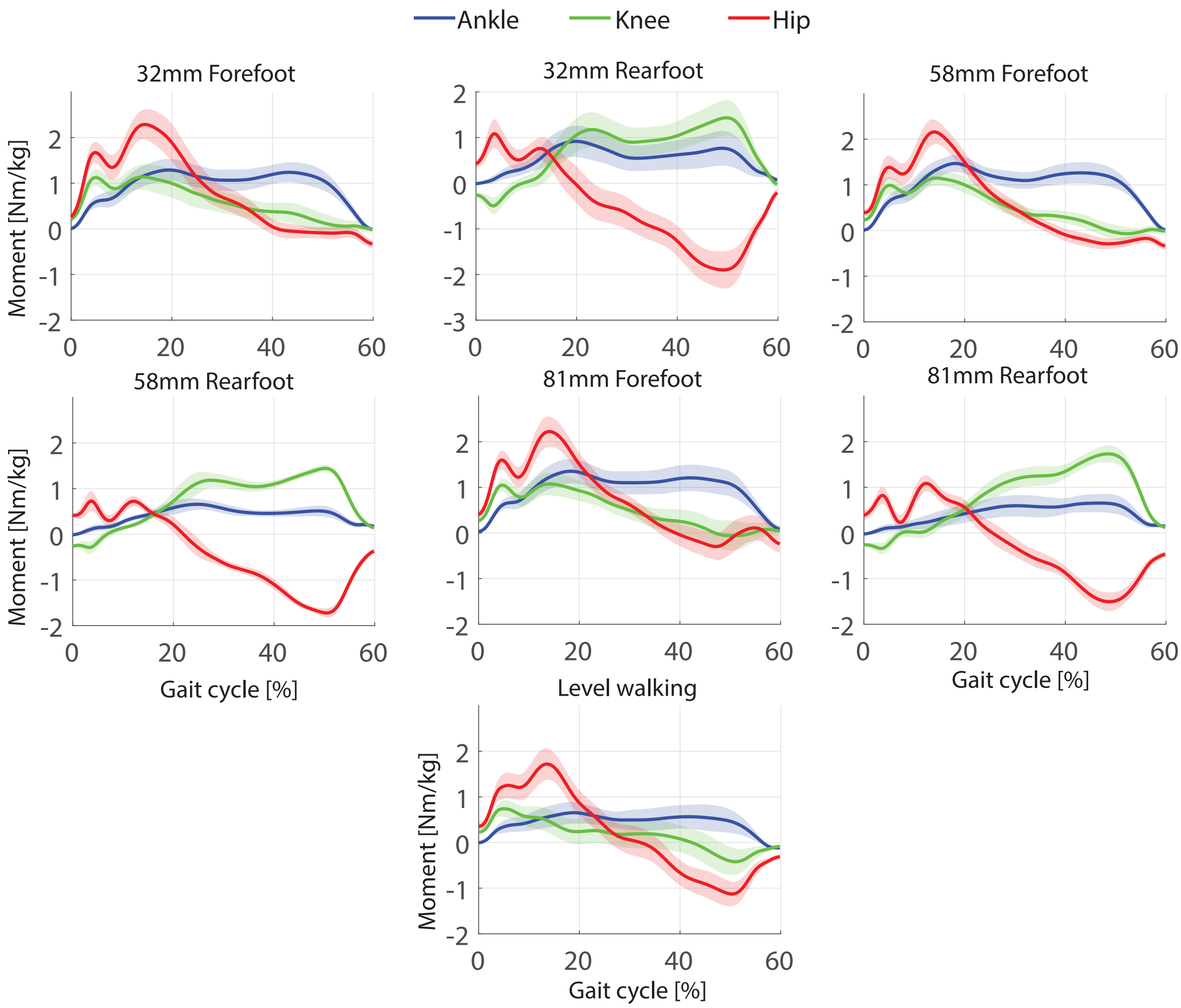

Figure 13: The figures describe the joint moment of the ankle, knee and hip for each condition separately during the stance phase (0% to 60% of gait). The shaded areas indicate the standard deviation. Positive moments indicate extension (plantarflexion at the ankle) and negative moments indicate flexion (dorsiflexion at the ankle).

D. Step length and step time

The figure presented below shows the results for the step length and time averaged over all subjects for each condition. No trend was found in the conditions significantly different from level walking.

Table 1: Step length and step time before (BB) and after (AB) beam hit. The values are averaged of all subjects for each trial. *Significantly different from level walking ($p < 0.05$). FF, forefoot; MF, midfoot; RF, rearfoot.

	Step length BB (SD) [m]	Step length AB (SD) [m]	Step time BB (SD) [s]	Step time AB (SD) [s]
Level	1.41 (0.05)	1.41 (0.05)	1.18 (0.03)	1.18 (0.03)
32mm 30deg FF	1.32 (0.23)	1.37 (0.22)	1.28 (0.09)	1.30 (0.18)
32mm 30deg RF	1.34 (0.19)	1.33 (0.13)	1.24 (0.07)	1.22 (0.15)
58mm 30deg FF	1.38 (0.16)	1.36 (0.15)	1.29 (0.08)	1.23 (0.08)
58mm 30deg RF	1.47 (0.16)	1.24 (0.13)*	1.35 (0.11)*	1.23 (0.08)
32mm FF	1.36 (0.15)	1.35 (0.07)	1.25 (0.11)	1.20 (0.09)
32mm MF	1.30 (0.14)*	1.32 (0.09)*	1.28 (0.07)*	1.28 (0.08)*
32mm RF	1.41 (0.12)	1.37 (0.08)	1.30 (0.07)*	1.23 (0.06)
58mm FF	1.43 (0.12)	1.42 (0.09)	1.34 (0.10)*	1.26 (0.07)
58mm MF	1.32 (0.15)	1.34 (0.16)	1.33 (0.08)*	1.29 (0.09)*
58mm RF	1.40 (0.19)	1.30 (0.20)	1.27 (0.09)	1.25 (0.14)
81mm FF	1.37 (0.09)	1.39 (0.07)	1.27 (0.09)	1.17 (0.03)
81mm MF	1.31 (0.12)	1.40 (0.04)	1.24 (0.07)	1.16 (0.03)
81mm RF	1.42 (0.08)	1.45 (0.03)	1.26 (0.10)	1.13 (0.03)