Faculteit Geneeskunde en Levenswetenschappen

master in de revalidatiewetenschappen en de kinesitherapie

Masterthesis

The Windlass Mechanism: Reliability of 2D video-analysis assessment methods

Sander Maes
Vic Wouters

Scriptie ingediend tot het behalen van de graad van master in de revalidatiewetenschappen en de kinesitherapie, afstudeerrichting revalidatiewetenschappen en kinesitherapie bij musculoskeletale aandoeningen

PROMOTOR :
dr. Pieter VAN NOTEN

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Acknowledgement

This master's thesis was written to obtain a Master's degree of Rehabilitation Sciences and Physiotherapy at the University of Hasselt.

We would like to express our profound gratitude to our promotor dr. Pieter Van Noten, for guiding and helping us during our research project. Despite the topic of our master's thesis falls outside of his expertise domain, he has taken up the challenge to investigate this unknown domain with us and regularly put us on the right track over the past two years throughout the whole process of our research. He has been helping us since day one of this master's thesis providing support, advice and guidance whenever asked for it. We want to specifically thank him for the help he provided with the statistical analysis of our data, a period in which he made time for us despite not having enough for himself.

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Also, we would like to thank prof. dr. K. Deschamps for the advice and guidance he provided in setting up our research protocol and for his willingness to answer any questions we had.

Our gratitude goes to REVAL and Adlon, to give us the possibility to do our research using their facilities and to dispose the materials needed.

We want to thank all the participants of this study who despite having nothing to gain from participating, took time out of their busy schedules to help us in acquiring data for our thesis.

Lastly, we would like to express our profound gratefulness to our family and friends, who have always supported us.

Hechtel-Eksel, 5 juni 2018 S.M.

Diest, 5 juni 2018 V.W.

1. Research context

This second part of our master's thesis falls under the domain of musculoskeletal rehabilitation, part of the study program of Rehabilitation Science and Physiotherapy at the University of Hasselt. This part will be an observational study in which we will examine the reliability of a video-analysis method to evaluate the function of the windlass mechanism of the foot. In contrast to the first part of our thesis, which was a literature study, this second part consists of a full fundamental research with a scientific output as aim. This study was conducted during the academic year of 2017-2018 by students Sander Maes and Vic Wouters under the supervision of promotor Dr. Pieter Van Noten. This master's thesis was a single study which was not part of a larger study or an already running project.

The windlass mechanism is an essential part of foot biomechanics which involves the tightening of the plantar fascia to provide stability and propulsion during gait. In our study we will examine the effect of this tightening on predominantly the medial longitudinal arch (MLA) of the foot by means of 2D video-analysis.

From our literature study we concluded that very few articles are available and objective data is scarce regarding the windlass mechanism. This makes it impossible to evaluate its function and/or dysfunction, so the need to know what windlass function is remains. Several tests have been developed, such as the windlass test for plantar fasciitis which is very specific but not sensitive (De Garceau, Dean, Requejo, & Thordarson, 2003), and cannot assess function objectively. The gold standard for evaluating foot biomechanics is 3D-analysis, but is not feasible for our thesis.

A correlation between static and dynamic settings will be examined, being a relation between a weight bearing (WB) and non-weight bearing (NWB) windlass test compared to normal gait. Magnitudes of MLA-movement, first metatarsophalangeal joint dorsiflexion (MTP1 DF) and rearfoot motion (RF) will be evaluated. The dynamic tests consisted of treadmill walking at preferred step rate and normal walking at that same step rate.

Since our goal was to develop a method to quickly and easily assess and evaluate windlass function, a 3D-analysis was not an option. The goal of this study was to examine the reliability of 2D-video analysis using relatively cheap methods, to warrant its use in a clinical setting. Both inter-rater and intra-rater reliability will be tested.

This topic arose from our own interest in foot biomechanics. The research design and methods were created by the students independently, under supervision and in consultation with promotor dr. Van Noten, and is based on previous research. In addition, the research design was discussed with prof. dr. Kevin Deschamps, who gave us tips and insight on how to improve our research protocol. The recruitment of subjects and the data-acquisition were performed by the students themselves, as well as the data-analysis. The statistics were discussed with and directed by dr. Pieter Van Noten. The academic writing of this thesis is independently done by the students, feedback was given by the promotor at the end of the process.

2. Abstract

Background: Research about the windlass mechanism is slowly on the rise. However, there is no consistency in current studies. There seems to be a need for some sort of standardised testing of normative values, since current evaluation of the windlass mechanism seems subjective.

Objectives: To develop and test a cheap and easy but reliable method to test windlass mechanism functioning, to warrant its use in a clinical setting.

Participants: Eighteen people participated in this study, of which thirteen were used in the analysis (mean age 34.8yrs, ranged 21-58). Two people were excluded due to a hallux limitus or hallux valgus, and three due to a loss of follow-up measurement.

Methods: 2D video-analysis of both dynamic (treadmill & normal surface walking/NS) and static tests (weight bearing (WB) & non-weight bearing (NWB) standing windlass test) was used. All tests were performed twice (pre- and post-test) and afterwards analysed by both examiners using Dartfish 9 Pro software. Statistical differences of results are calculated using JMP Pro 13.2 as well the ICC (Intra-Class Correlation coefficient) to assess reliability of the tests.

Results: Differences in MLA could be seen comparing initial contact (IC), midstance (MS) and toe-off (TO) values in the treadmill and normal walking test. The results were the same for dorsiflexion angles and rearfoot motion. In the static tests a difference could be seen for MLA-angle comparing weight bearing to non-weight bearing, but not for dorsiflexion. When comparing the magnitude of MLA range of motion (ROM) between static tests (WB & NWB) and normal walking, MLA ROM in normal walking was almost twice as much (12.0717° vs 6.5148° & 4.74193°, $p<0.0001$*). ICC-values ranged from excellent to poor. Inter-rater reliability for the treadmill tests was excellent (ICC>0.90). Good results were found for the intra-rater reliability for the treadmill test for MLA, inter-rater for DF-angle and inter-rater for static tests for DF-angle (ICC>0.70). Moderate results were found for intra-rater reliability for the treadmill test for DF and RF and inter-rater in normal walking tests (ICC>0.55). Poor results were found for inter-rater in treadmill tests for RF-angles and static tests for MLA-angles (ICC<0.40).

Conclusion: ICC-values show that the treadmill test is very reliable for assessing MLA-angles, and moderately reliable for MLA using the normal walking test. No significant differences were found in MLA-angle comparing the treadmill test to the normal walking test.

Using these easy methods, a fairly reliable assessment of MLA-angle can be made during walking. Static tests cannot be used to assess dynamic function. Future research should focus on the dynamic relation between MTP1 dorsiflexion and MLA-angle changes, and the kinematic relation between windlass mechanism and the lower limb.

3. Introduction

The windlass mechanism of the foot is an essential aspect of foot biomechanics and is crucial during the gait pattern. It was first described by Hicks in 1954 as the winding up of the plantar fascia when the toes move to a dorsiflexed position. This pulls on the fascia which increases tension (Carlson, Fleming, & Hutton, 2000). This tension is then transferred to both the calcaneus and the base of the first metatarsal which move closer to each other because of that increased tension in the plantar fascia. The movement of these two bones must be compensated somewhere, so as a result the entire medial longitudinal arch (MLA) increases in height, and the calcaneus supinates due to the place of insertion of the plantar fascia on the medial calcaneal tubercle. The raising of the MLA predominantly happens in the naviculo-cuneiform joint. (Hicks, 1954; Fuller, 2000; Caravaggi, Pataky, Gunther, Savage, & Crompton, 2010; Bolgla & Malone, 2004). This mechanism provides stability during normal gait, as a well-functioning windlass effect limits the amount of rearfoot eversion during the stance phase, limiting excessive pronation and thus creating a more stable arch (Kappel-Bargas, Woolf, Cornwall, & McPoil, 1998; Fuller, 2000; Nakamura & Kakurai, 2003). There is however a lack of some objective standard or normative values to evaluate windlass functioning or dysfunctioning. Subsequently, we cannot accurately define a windlass function or dysfunction. Several studies have attempted to examine windlass function and some theories have been developed.

A first way to describe function is to look at the onset time of the arch-rising motion when dorsiflexing the toes. Kappel-Bargas et al. (1998) conducted an experiment where the first metatarsophalangeal joint (MTP1) was passively dorsiflexed and the MLA-movement was recorded. They found that the sample population (n=20) could be divided into two groups: an immediate onset group and a delayed onset group. In the immediate onset group, the MLA started to rise after 4.1° of MTP1 dorsiflexion. In the delayed onset group MLA-rise started at 20.4° dorsiflexion. Based on these findings, a possible way to classify different types of windlass functioning can be described (Kappel-Bargas et al., 1998). Nakamura and Kakurai (2003) proposed another way to assess windlass function. The findings of their study suggest that a classification based on the time to maximum rearfoot eversion can be made, expressed as a percentage of the stance phase. They defined two groups as ‘early eversion onset’ and ‘late eversion onset’. Between these two groups several significant differences were found, such as

the time to maximum rearfoot eversion, expressed in a percentage of the stance phase and with the late eversion onset group reaching maximum rearfoot eversion later at 71.1% vs 41.8% of stance phase. Other significant differences were the degree of maximum rearfoot eversion with 3.74° vs 7.69°, maximum MLA-angle change with 4.7° vs 7.24° and time to maximum MTP1 extension with 41.2% vs 64.3% of stance phase, for early eversion onset and late eversion onset respectively. These variables could also lead to a possible way to define windlass function. A recent study by Lucas and Cornwall (2017) investigated the static foot posture of feet that had a functioning, impaired or absent windlass mechanism. This classification was based purely on visual establishment of the windlass mechanism after passive MTP1 dorsiflexion, in a standing position. Lucas and Cornwall described their method as the 'Big Toe Extension Test', which is a static test. They found a difference in the Foot Posture Index (FPI) -which is a cluster of eight static foot posture measurements- when comparing these groups, with the 'Absent' group scoring lower than the 'Functioning' group (Redmond, Crosbie, & Ouvrier, 2006; Lucas & Cornwall, 2017). An earlier study by Aquino and Payne (2001) however found that static foot posture is not linked to dynamic windlass function, so the importance of the FPI in relation to dynamic windlass function during walking may not be very great. Lucas and Cornwall stated in their study that more research is necessary to investigate the relation between MTP1 DF and MLA changes in dynamic situations. This study partly delivers that.
To date, these are the only studies who have attempted to objectify and classify different types of windlass mechanism functioning. There have not been any follow-up or more in-depth studies confirming or contradicting these results. Therefore, it remains difficult to objectively define dynamic windlass function, or more importantly, dysfunction. When there is no definition of windlass function we cannot possibly define a malfunctioning, or devise a way to assess it.

An objective standard for windlass dysfunction does not exist. Nor is it known what a normal response of the MLA or rearfoot to a dorsiflexion of the toes is. In this study, the relation between MLA-changes and DF-changes during walking will be examined. Hicks (1954) noted that the first ray has approximately 22° of ROM for arch height increase and decrease measured in the naviculo-cuneiform joint and that about 50% (10°) of that ROM could be attributed to the windlass mechanism. Tansey and Briggs did a study in 2001 where the contribution of the windlass mechanism to heel supination was measured. They concluded that about 50% of heel supination was due to the windlass mechanism, while the other 50% was due to active mechanisms (i.e. muscular components). (Tansey & Briggs, 2001). Thordarson, Kumar, Hedman, and Ebramzadeh (1997) also investigated the effect of the plantar fascia in cadavers on MLA-height increase. They measured the arch height in both 30° and full MTP1 extension with intact plantar fasciae, followed by the same measurements when cutting the plantar fascia in one quarter increments. With each quarter arch height decreased. Ultimately after a complete fasciotomy, a 50% decrease of arch height with complete fasciotomy versus an intact plantar fascia was found. This is in line with the findings of Tansey and Briggs which suggests the windlass mechanism is responsible for approximately 50% of motion in the MLA and the rearfoot. These hypotheses will also be tested in this study.

There are several theories that windlass mechanism dysfunction can attribute to lower limb injury, with the most common one being a relation with plantar fasciitis. However, relationships with the achilles and patellar tendons have also been proposed. From a biomechanical perspective the idea is that the windlass mechanism can affect the positions of the lower limbs. The windlass mechanism supinates the rearfoot, or calcaneum which is the insertion of the achilles tendon. This supination is also paired with an external rotation of the tibia, to which the patellar tendon inserts. (Fuller, 2000; Prior, 1999). A biomechanical relation is clear, however there is lack of evidence. A very recent study by Manfredi-Márquez et al. (2017) attempted to measure movement throughout the entire lower limb as caused by the windlass mechanism. It was clear from their results that the windlass mechanism can cause movement throughout the entire lower limb, with more motion in the more distal segments and lessening when measuring more proximally. This suggest a possible role of the windlass mechanism in lower limb injury. Their measurement system was however flawed, as the sensors used were very large and caused movement of the skin over bony segments of the foot (Manfredi-Márquez et al., 2017). Therefore, in this study, plasticised paper markers will be used to mostly eliminate this bias.

Clinical tests to evaluate windlass mechanism functioning are scarce. The Windlass Test is the only clinical test to evaluate windlass function, but relies purely on a visual assessment of windlass establishment. When passively dorsiflexing the big toe, the examiner should look at a potential arch-height increase and a band-like protrusion of the plantar fascia on the plantar aspect of the foot. These two aspects are typically used to define 'windlass establishment'. This test however has only been evaluated for plantar fasciitis by De Garceau et al. (2003), who found a specificity of 100%, but a sensitivity of 32% when testing for plantar fasciitis. It can therefore not be used reliably.

A good alternative which is accessible to everyone in a modern-day clinic is 2D video-analysis. No expensive equipment, tests, lengthy analysis or third parties need to be involved. Every clinician with a video camera capable of slow-motion capture (at least 100fps) can perform such an analysis. Therefore, in this master's thesis, the usefulness of 2D-video analysis for windlass function using everyday equipment will be investigated. This way, a possible method can be devised for everyday practice which is less time-consuming, and close to equally accurate as 3D-analysis.

In conclusion, to date no clear and easy method of assessing windlass function has been established. Several tests are available such as 3D analysis and the windlass test, but none being easy, quick and accurate all at the same time. Therefore, the goal of this study was to evaluate the reliability of 2D video-analysis using relatively cheap methods to potentially warrant use for clinical settings, and to possibly establish some normative values for windlass mechanism functioning.

4. Methods

4.1. Subjects

Eighteen subjects participated in this study (9 males, 9 females). A questionnaire was filled out by everyone before participating to identify possible exclusion criteria (see Appendix A for the questionnaire). Subjects were excluded if they had an injury or surgery to the lower limb within the past 6 months, had diabetes or had a neurological disorder that affected their motor control and gait pattern. Furthermore, subjects were excluded if they could have altered foot biomechanics due to conditions such as hallux limitus/rigidus and hallux valgus. Healthy individuals with an age between 18 and 65 years were included. Eventually, thirteen of the originally eighteen were analysed (6 males, 7 females). Two subjects were excluded due to a hallux limitus or hallux valgus which were only noticed during testing, and three due to a loss of follow-up measurement.

This study (code: B9115201734634) was reviewed and approved by the Board of Ethics at Hasselt University on the 20th of December 2017 and every participant was given and signed an informed consent form.

4.2. Procedures

Our protocol consisted of several parts. First, subjects were measured for height and weight, as well as their foot length and truncated foot length. Next, tracking-markers (provided by Dartfish) were taped using hypoallergenic tape on anatomical landmarks on the subject's feet. For the MLA-angle, markers were placed at the tuberculum naviculare, the MTP1 joint and the medial tuberculum of the calcaneum. For the DF-angle, the markers at the MTP1 joint and tuberculum naviculare were also used, and one extra at the most medial aspect of the distal phalanx of the hallux (see Appendix B for placement of the markers).

Thereafter, subjects were asked to walk on a treadmill at a comfortable pace which they could adjust themselves. Once a comfortable pace was found for one minute, three cameras recorded the treadmill walking for one minute; two in the sagittal plane (left and right of the treadmill), and one posterior to record movement in the frontal plane. The subjects kept walking while the examiner determined their step frequency using a metronome. The position of the cameras was fixed during the pre- and post-measurement.

For the next part, subjects walked normally on the ground in a figure-of-eight (Normal Surface Walking/NSW). The long parts of this figure-of-eight were approximately 8 meters in length. In the middle two cameras were placed laterally of the walkway, so that the medial side of both feet could be recorded. With the help of a metronome, subjects were asked to walk at the same step frequency they had walked on the treadmill, and had to keep walking until a minimum of 5 steps were recorded within the camera frame.

In both the treadmill walking and normal surface walking, three moments in the stance phase were recorded. First at initial contact (IC), being the moment the heel touches the ground. Next at midstance (MS), which in this study was the moment right before the swinging foot passes in front of the standing foot. Lastly, the moment at which the DF-angle of the hallux was smallest, was recorded. In this study this moment will be labelled as toe-off (TO).

Finally, the static test was evaluated by standing on a scale with one foot, with the MTP1J right at the edge of the scale, so that the hallux was hanging over. In this position, participants were asked to put 10% or less of their bodyweight on the scale (non-weight bearing/NWB), and afterwards 90% or more of their bodyweight on the scale (weight bearing/WB). In these two positions the hallux was passively dorsiflexed maximally three times, and then repeated with the other foot. A camera recorded the medial side of the foot.

All tests were repeated entirely at the follow-up, except for height, weight and foot length.

Because the reliability of these tests was to be investigated, all subjects were measured twice on separate days, with a minimum of three days in between measurements and a maximum of 14 days (See Appendix C for images of the testing setup).

4.3. Data-analysis

4.3.1 Video-analysis

All videos were analysed using Dartfish 9 Pro. All the data obtained in the Dartfish software was imported to Excel files. The data was analysed separately by the two examiners, each examiner determined the MLA-angle, DF-angle and RF-angle of five steps of the right and left foot, for both the pre- and post-measurements for the treadmill test. For the NSW and WB/NWB, semi-automatic tracking was used to determine MLA-angle and DF-angle. In the line of reasoning a rater effect would be minimal when semi-automatic tracking is used, it was decided not to do an inter-rater analysis for the NSW and WB/NWB tests. Only an inter-rater analysis was performed for the manual placement of the MLA- and DF-angles on the treadmill tests. For the NSW test, rater 1 determined MLA- and DF-angles of five steps of the right and left foot using semi-automated tracking for the pre-tests, rater 2 did the same for the post-test. For the WB and NWB tests, rater 1 analysed all three trials of the pre-test, while rater 2 analysed the three trials of the post-test.

4.3.2 Statistical analysis

Data was analysed using JMP Pro 13.2 software. First, normality and homoscedasticity of the data was assessed using the Shapiro-Wilk Wand test and the O'Brien test respectively.

All treadmill-data, except for MS DF (PRE), IC DF (PRE) and MS RF (POST), were not normally distributed. All NSW-data were normally distributed, except for MS DF (POST). The data of the static WB and NWB tests were all normally distributed.

Homoscedasticity was also checked, a significant difference ($p<0.05$) was found for several datasets, so data were not consistently homoscedastic.

Because of not meeting the conditions of normality and homoscedasticity, it was decided to use non-parametric statistical tests to analyse all data.

Matched pairs analyses were used to compare data from all tests. The non-parametric Wilcoxon signed-rank test was used for this analysis. For the treadmill test, to test pre-post differences, the data of the pre-test at IC was paired with IC of the post-test, as well as for MS and TO. Depending on which analysis was performed (pre-post/intra-rater/inter-rater) different pairs were made in all other tests as well. For example, to test inter-rater differences, the data for IC, MS an TO from different raters were paired instead of the pre- with the post-test data.

Main effects were also tested using specialised modelling. When a main effect was found, post-hoc testing was performed using the Steel-Dwass test to determine true significance. Alpha levels were set at 0.05.

The Intra-Class Correlation (ICC) coefficients were also calculated for each angle and for each phase of the gait to test the reliability of the methods used (see Appendix D for the results of the ICC).

To assess for a correlation between the changes of the MLA- and DF-angles, a Spearman's Rho was calculated using the data from the normal surface walking test.

5. Results

5.1. Treadmill

Results of the treadmill tests can be seen in Figure 1, 2 and 3. To eliminate bias, the difference in step frequencies was also tested. There was no significant difference in step frequency between the pre-test and the post-test ($p=0.7148$).

Differences were found for every comparison (IC-MS, IC-TO, MS-TO) for every rater ($p<0.0001$*). From IC to MS, the MLA increased with 5.71°, after which it decreased again from MS to TO by 10.33° The difference between IC and TO was 4.62°. The DF-angle decreased by 18.93° from IC to MS and then decreased by 41.56° from MS to TO, with the difference between IC and TO being 22.63°. Lastly the RF-angle changed by 4.33° of pronation from IC to MS, then supinated by 8.50° from MS to TO, and differed by 4.17° between IC to TO. In figure 3, negative values mean a supinated position of the rearfoot, whereas positive values indicate a pronated rearfoot.

The intra-rater reliability for the treadmill data was also tested. Data was tested for each outcome measured separately (MLA-DF-RF). For rater 1, significant differences were found in the DF at midstance ($p=0.0004$*) and toe-off ($p=0.0006$*), and for the RF-angle at all three phases ($p<0.05$*). For rater 2, significant differences were found for the DF-angle at initial contact ($p=0.0074$*) and toe-off ($p=0.0009$*) and for the RF-angle at midstance only ($p<0.0001$*). Both raters had significantly different results comparing the pre- and post-treadmill test for DF at TO and RF at MS. No intra-rater differences were found for the MLA-angle. All ICC-coefficients ranged between 0.50535-0.7793. No main effect was found in the analysis.

Inter-rater reliability was also evaluated. For the pre-test, significant differences were found for the DF-angle at midstance and toe-off ($p<0.0001$*) as well as for RF-angle at all three phases ($p<0.01$*). There was a rater effect for MLA at TO, DF at MS and TO, and RF at IC and TO. Post-hoc analyses found that all but RF-angles at TO ($p=0.9686$) were truly significantly different ($p<0.05$*). In the post-test, significant inter-rater differences were found for MLA-angle at initial contact ($p<0.0001$*) and midstance ($p=0.0323$*), for DF-angle at midstance and toe-off ($p<0.0001$*), and for RF-angle at all three phases ($p<0.001$*). ICC-analysis results were similar. All MLA values had ICC-coefficient of 0.9 or higher. For DF-angles the ICC-values ranged between 0.5686-0.8055. For RF-angles ICC-values were lowest, ranging between 0.2694-0.5125.

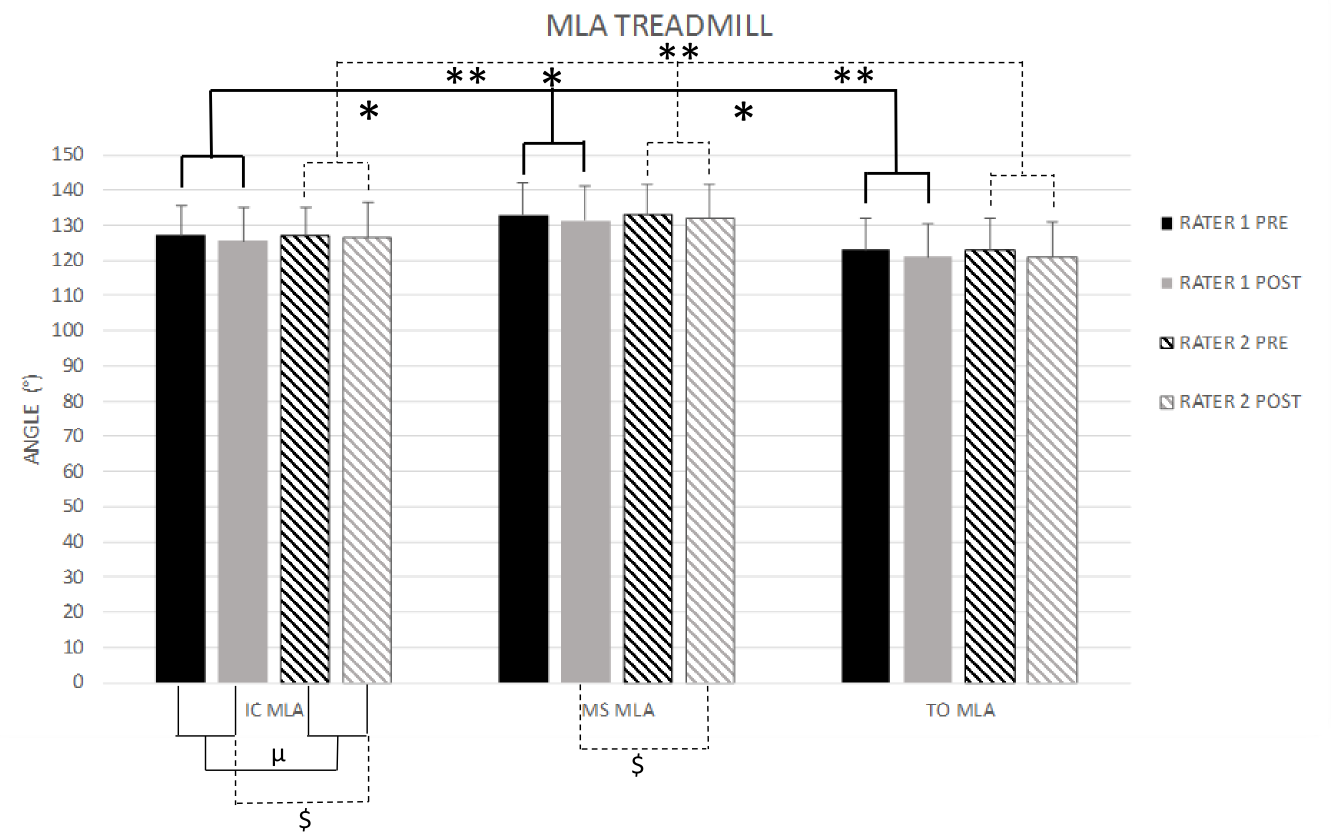


Figure 1: Mean MLA-angle at IC, MS and TO for rater 1 and rater 2 for both pre- and post-test in the treadmill test.

*: Rater 1 combined pre- and post-test significant difference between IC, MS and TO MLA-angle (p-value<0,05).

**: Rater 2 combined pre- and post-test significant difference between IC, MS and TO MLA-angle (p-value<0,05).

µ: Inter-rater (combined pre- and post-test) significant difference for IC MLA (p-value<0,05).

$: Inter-rater post-test significant difference for IC MLA and MS MLA (p-value<0,05).

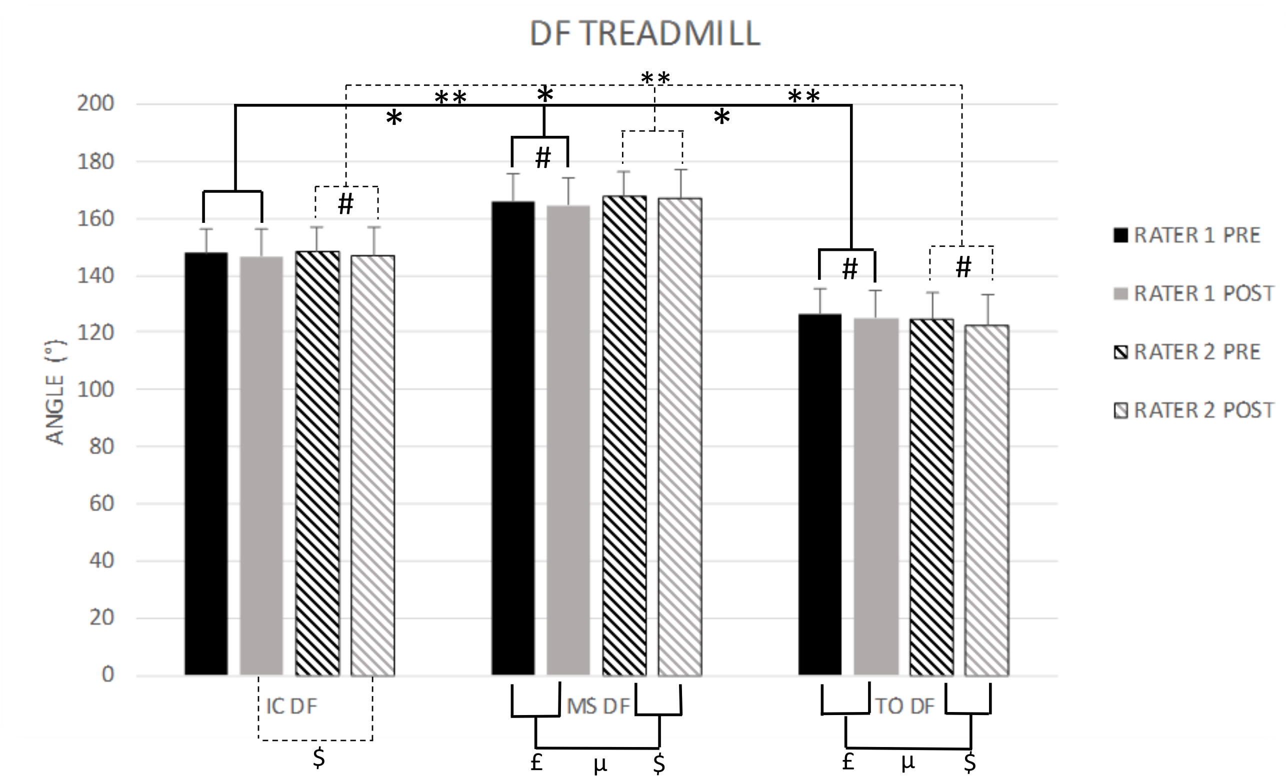


Figure 2: Mean DF-angle at IC, MS and TO for rater 1 and rater 2 for both pre- and post-test in the treadmill test.

*: Rater 1 combined pre- and post-test significant difference between IC, MS and TO DF-angle (p-value<0,05).

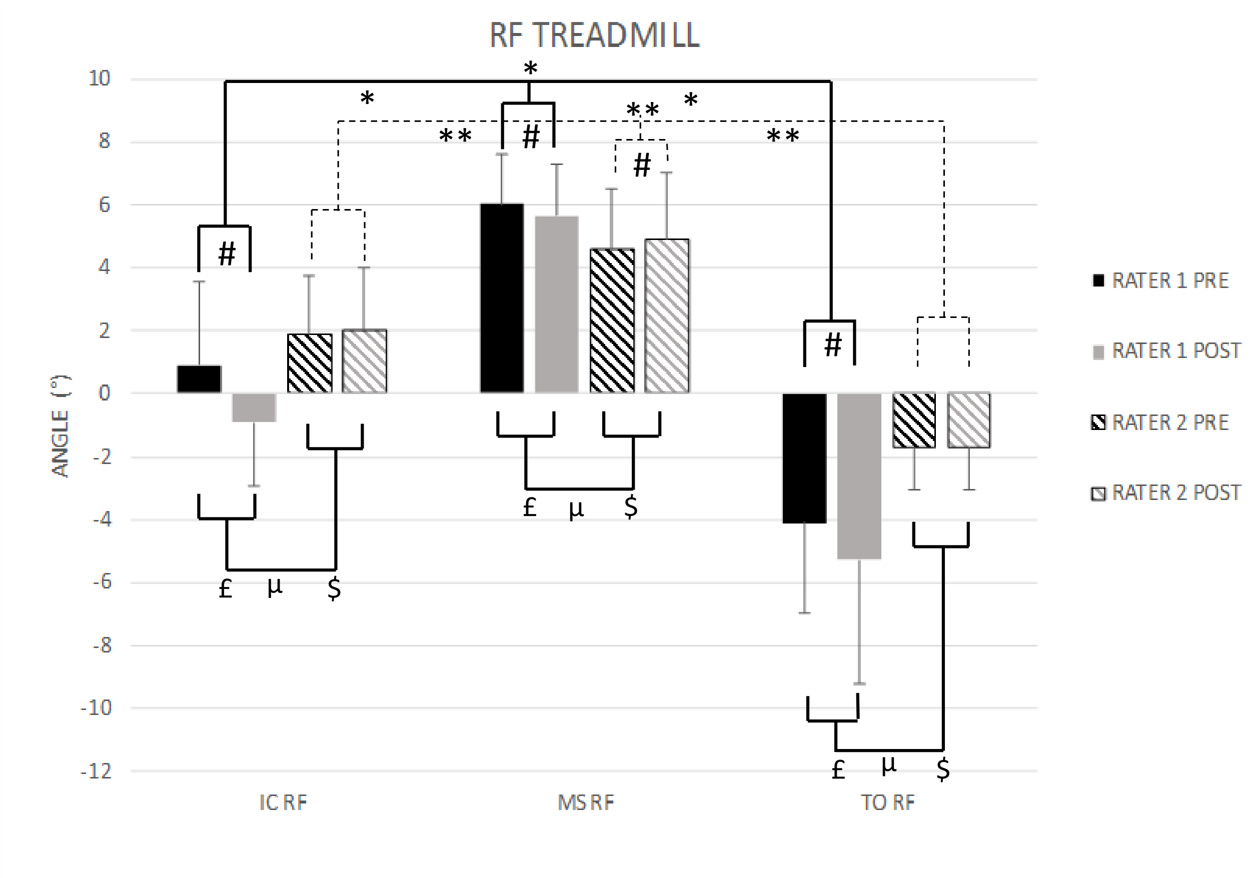
**: Rater 2 combined pre- and post-test significant difference between IC, MS and TO DF-angle (p-value<0,05).

µ: Inter-rater (combined pre- and post-test) significant difference for MS and TO DF-angle (p-value<0,05).

$: Inter-rater post-test significant difference for IC, MS and TO DF-angle (p-value<0,05).

£: Inter-rater pre-test significant difference for MS and TO DF-angle (p-value<0,05).

#: Intra-rater pre-post significant difference for MS and TO DF-angle Rater 1 and IC and TO DF-angle Rater 2 (p-value<0,05).



<u>Figure 3</u>: Mean RF-angle at IC, MS and TO for rater 1 and rater 2 for both pre- and post-test in the treadmill test.

*: Rater 1 combined pre- and post-test significant difference between IC, MS and TO RF-angle (p-value<0,05).

**: Rater 2 combined pre- and post-test significant difference between IC, MS and TO RF-angle (p-value<0,05).

#: Intra-rater pre-post significant difference for IC, MS and TO RF-angle Rater 1 and MS RF-angle Rater 2 (p-value<0,05).

µ: Inter-rater (combined pre- and post-test) significant difference for IC, MS and TO RF-angle (p-value<0,05).

$: Inter-rater post-test significant difference for IC, MS and TO RF-angle (p-value<0,05).

£: Inter-rater pre-test significant difference for IC, MS and TO RF-angle (p-value<0,05).

5.2. Normal surface walking

For MLA, significant differences were found for all three phases for both pre- and post-tests (p<0.0001*), with mean differences of 4.107085°, 4.9636°, 9.0707° for MS-IC, TO-IC and TO-MS respectively (Figure 4). The same results were found for the DF-angle with mean differences of 12.783°, 25.683°, 38.466° for MS-IC, TO-IC and TO-MS respectively (p<0.0001*) (Figure 5). Furthermore, pre- and post-tests were compared for each moment and both MLA- and DF-angles. No differences were found comparing pre- and post-tests for either MLA or DF-angle, with mean differences ranging between 0.0427° and 1.3833°.

ICC-analysis yielded highly reliable results for MLA-angles (ICC 0.7467-0.8872), and moderately reliable for DF-angles (ICC 0.5992-0.7359).

A Spearman's Rho was calculated to test the relation between the MLA-angle and the DF-angle, which was found to be 0.37 (p<0.05).

A main effect analysis was run and only one significant main effect was found for the DF-angle at TO for pre-post testing. Post-hoc analysis showed a significant difference between pre-test and post-test for TO DF (p=0.0091*).

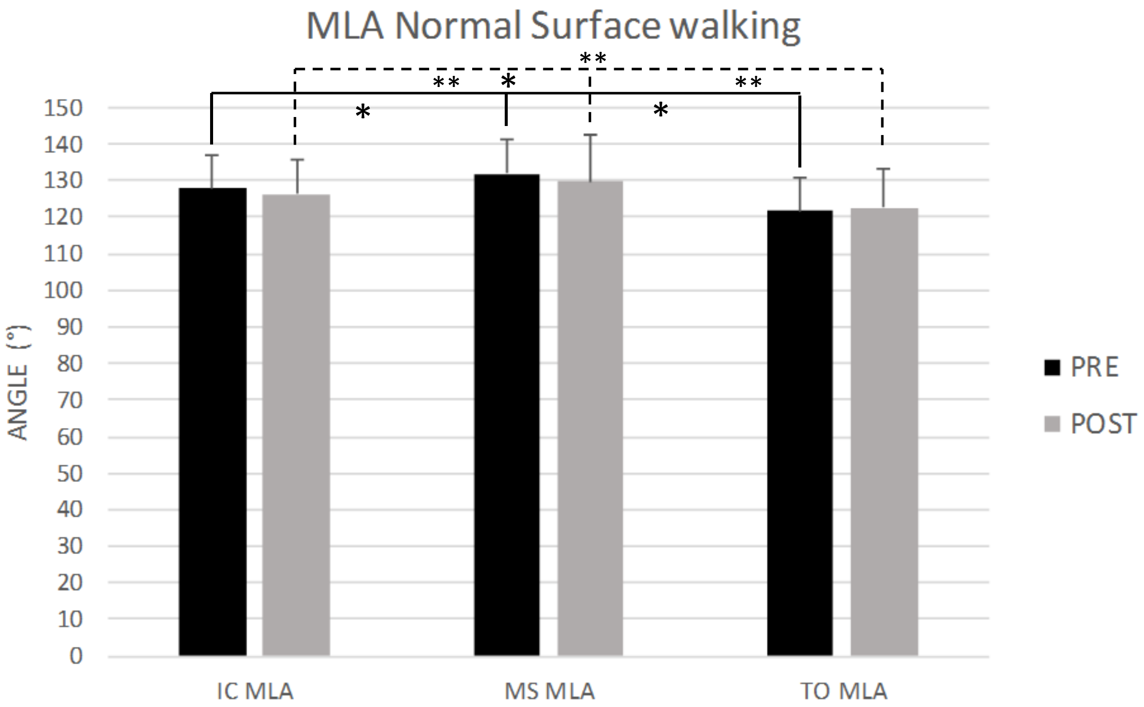


Figure 4: MLA-angles in normal surface walking.

*: Significant differences (p-value<0.05) between pre-IC, -MS and -TO MLA.

**: Significant differences (p-value<0.05) between post-IC, -MS and -TO MLA.

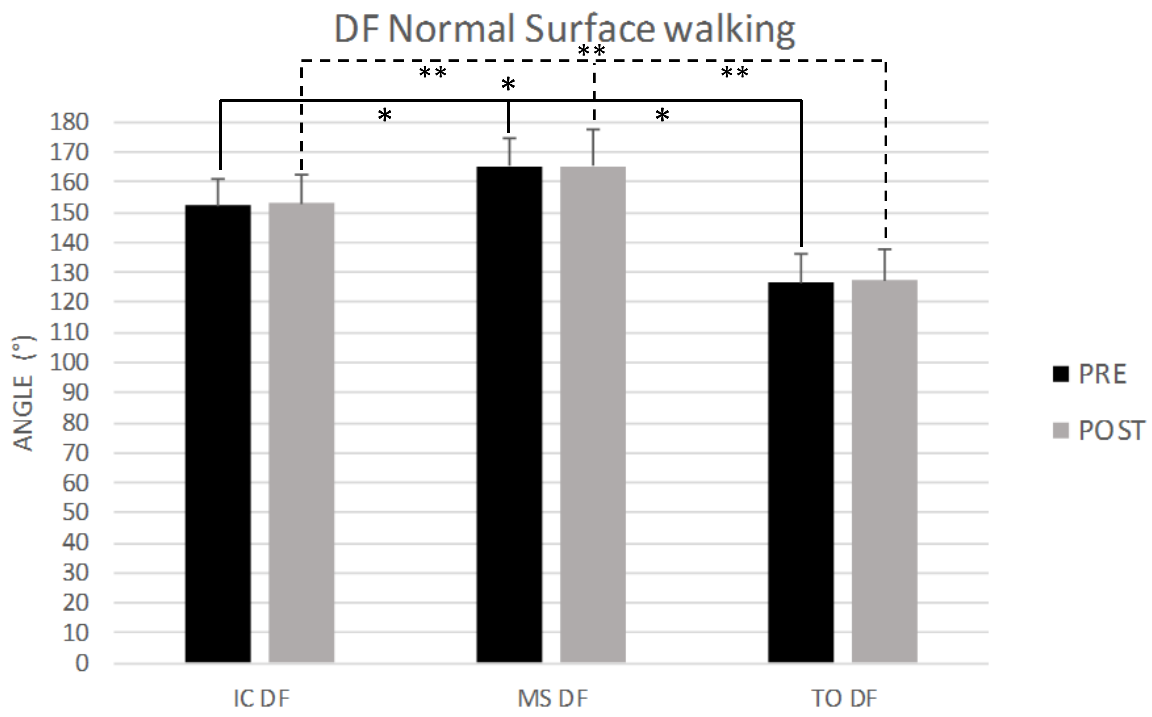


Figure 5: DF-angles in normal surface walking.

*: Significant differences (p-value<0.05) between pre-IC, -MS and -TO DF.

**: Significant differences (p-value<0.05) between post-IC, -MS and -TO DF.

5.3. Weight bearing and non-weight bearing

The first hypothesis to be tested were whether there was a difference in the magnitude of either MLA or DF angle between a weight bearing and non-weight bearing standing windlass test, with pre- and post-tests examined separately. To test this, the difference between the maximum and minimum angle recorded was calculated for each trial for each subject and used as the mean MLA difference to compare to.

A significant difference for MLA was found in the post-test with a mean difference of 2.067° ($p<0.0001$*), meaning in the post-test, the change in MLA was significantly different in the weight bearing versus the non-weight bearing test. For the DF-angle, no significant differences were found between weight bearing and non-weight bearing tests (mean diff: 1.41° and 0.243° for pre- and post- test respectively).

The difference between the pre- and post-test for both weight bearing and non-weight bearing was also examined. For the MLA-angle, a significant difference was found between pre- and post-tests for the non-weight bearing test ($p=0.0132$*, mean diff: 1.165°). For the DF-angle no significant differences were found.

ICC-tests found very poor correlations for the MLA-angle (ICC 0.183-0.3614) and excellent values for DF (ICC 0.8198-0.8382).

5.4. Treadmill vs normal surface walking

To compare the treadmill data to the data of the normal surface walking test, the pre- and post-data were combined into one mean value. MLA and DF-angles were compared between the two tests. For MLA-angle, mean differences were 1.1178°, 0.73696° and 1.2915° for IC, MS and TO respectively, but were not found to be significant. For the DF-angle in NS compared to treadmill, significant differences were found for IC (6.00°, $p<0.0001$*) and for TO (3.026°, $p=0.0014$*). The results can be seen in figure 6 and 7. The mean values of the treadmill and the normal surface walking data are displayed in table 1 (see Appendix E for table with means and standard deviation).

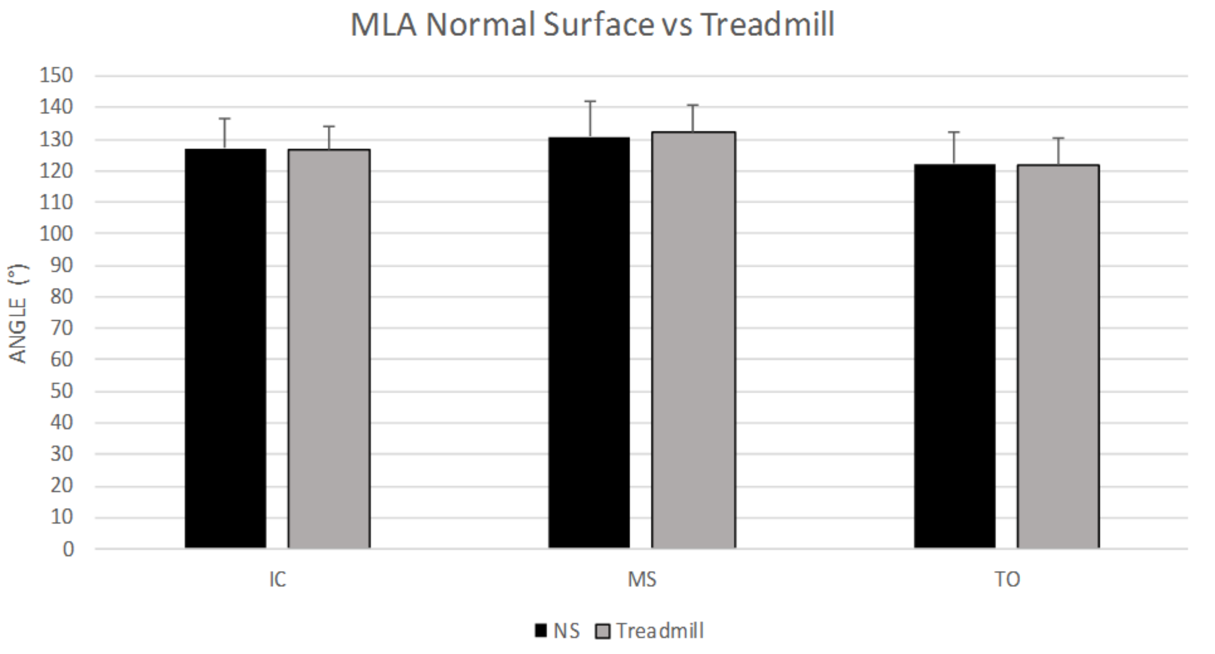


Figure 6: Differences in MLA-angle between NS and treadmill walking. No significant differences (p-value>0.05) were found for the MLA-angle during IC NS & IC treadmill, MS NS & MS Treadmill and TO NS & TO Treadmill.

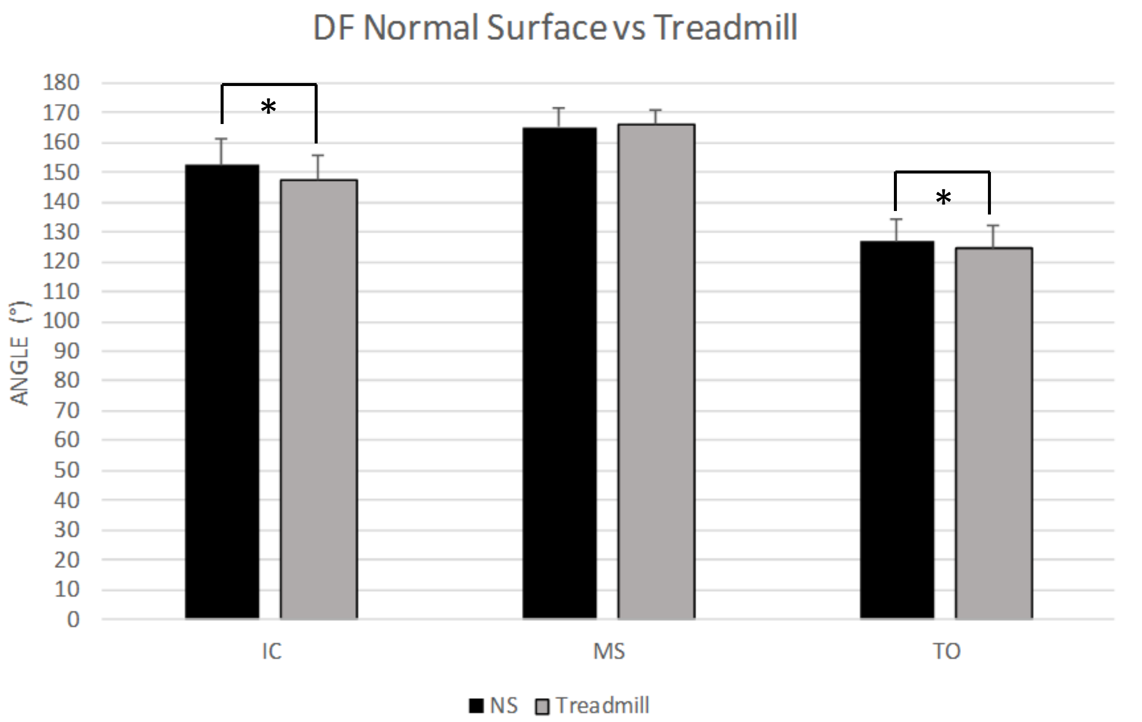


Figure 7: Differences in DF-angle between NS and treadmill walking.

*: Significant difference (p-value<0.05) was found for the DF-angle during IC NS & IC Treadmill and during TO NS & TO Treadmill.

5.5. Normal surface walking vs weight bearing and non-weight bearing

The magnitudes of ROM were compared between the WB/NWB and NS walking. The maximum and minimum angles were taken and the difference between the two was calculated. These values were used for comparison. The pre- and post-tests were evaluated separately.

In the pre-test, significant differences were found for the MLA-angle when comparing NSW with WB and NWB tests. In the pre-test mean differences were 5.80278° for NS-WB (12.516 vs 6.71323), and 7.28112° for NS-NWB (12.516 vs 5.23489) ($p<0.0001$*). The same was found in the post-tests with mean differences of 5.31105° (11.6274 vs 6.31636) and 7.37843° (11.6274 vs 4.24897) ($p<0.0001$°).

For the DF-angle, the same tests were all significantly different, with mean differences of 16.889°, 15.552° for NS-WB and 17.969°, 17.726° for NS-NWB for pre- and post-tests respectively ($p<0.0001$*). The results can be seen in figure 8 and 9. No main effects were found for the pre- and post-test.

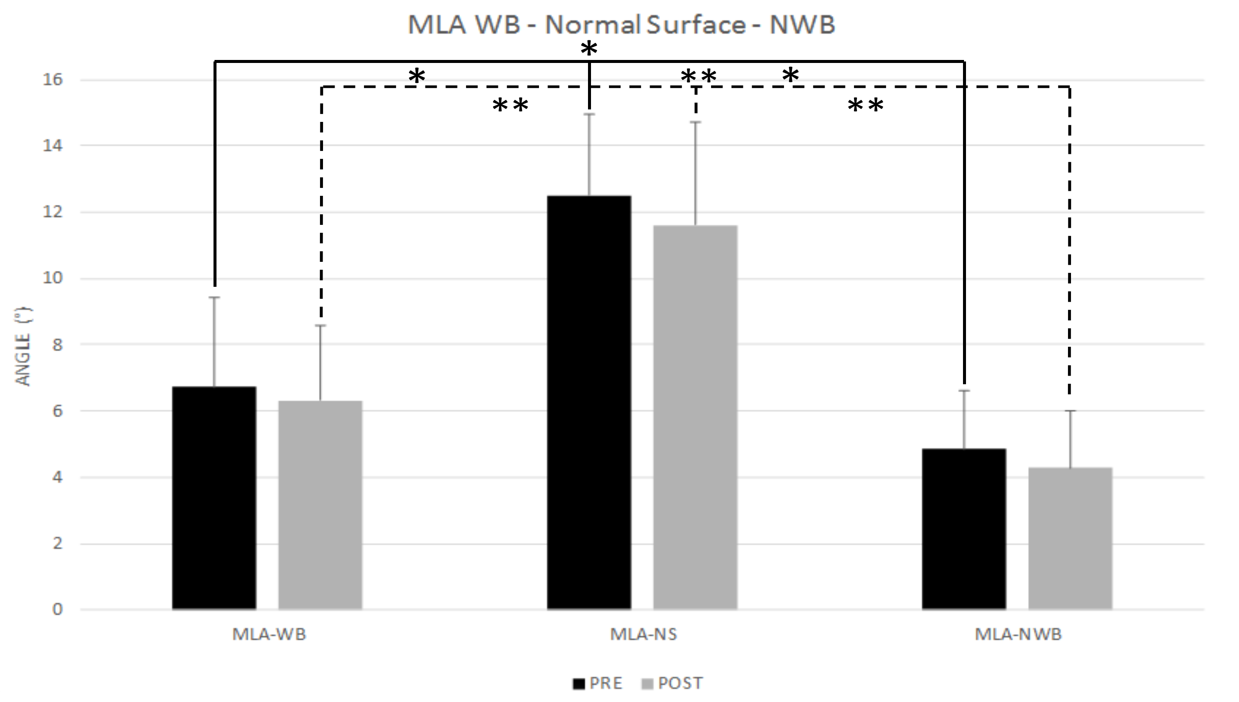


Figure 8: ΔMLA-angle between NS-WB-NWB tests.

*: Significant difference (p-value<0,05) between pre-test MLA-WB (weight bearing) and pre-test MLA-NS (normal surface walking), pre-test MLA-NWB (non-weight bearing) and pre-test MLA-NS, pre-test MLA-WB and pre-test MLA-NWB.

**: Significant difference (p-value<0,05) between post-test MLA-WB and post-test MLA-NS, post-test MLA-NWB and post-test MLA-NS, post-test MLA-WB and post-test MLA-NWB.

No significant pre-post differences for MLA-WB, MLA-NS and MLA-NWB.

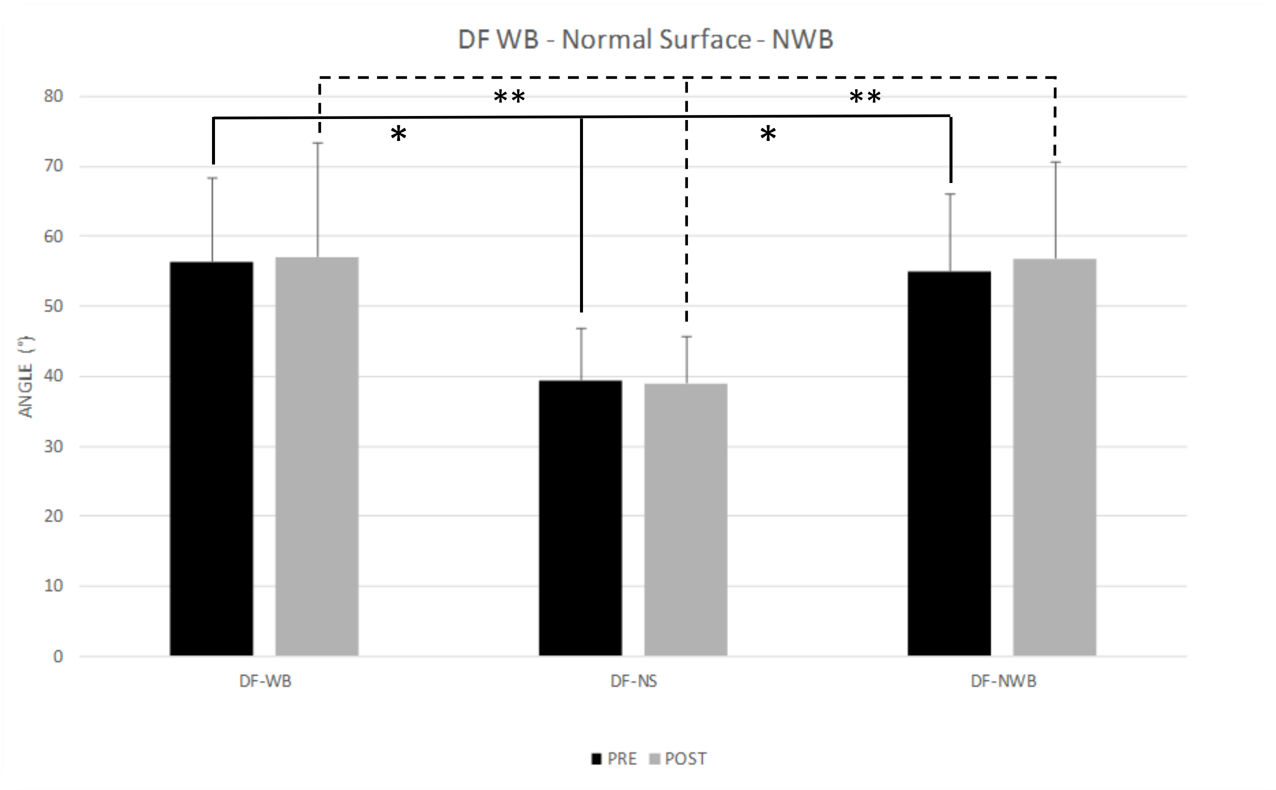


Figure 9: ΔDF-angle between NS-WB-NWB tests.

*: Significant difference (p-value<0,05) between pre-test DF-WB (weight bearing) and pre-test DF-NS (normal surface walking) & between pre-test DF-NWB (non-weight bearing) and pre-test DF-NS.

**: Significant difference (p-value<0,05) between post DF-WB and post-test DF-NS & between post-test DF-NWB and post DF-NS.

No significant pre-post difference for MLA-WB, MLA-NS and MLA-NWB.

No significant difference between DF-WB and DF NWB (both pre- and post-test).

6. Discussion

The most important findings of this study are the ICC-values for all tests. This was the main purpose of this study; to evaluate whether methods which are accessible in everyday practices can be used reliably. Another important aspect of this study is the comparison of three moments in the stance phase between static tests and dynamic situations. In the authors opinion, this would be the first study to compare data from dynamic tests to data from static tests. However, before such comparisons could be made, the authors wanted to prove that with their method, changes in MLA and DF could be observed and measured during the stance phase. Since two dynamic situations were tested (treadmill and normal surface), the authors thought it would be interesting to also compare the changes in MLA and DF between the two dynamic tests.

When looking at the walking pattern of both the treadmill and normal surface tests, both the MLA and DF-angles change during the stance phase. Figure 10 shows that smaller MLA angles relate with small DF-angles, and vice versa (see Appendix F for figure), although the relation is not very monotonic (Spearman's Rho= 0.37). This could be explained by the small lag in MLA-change following a DF-change. Later in the stance phase, it can be observed that the MLA lags slightly behind the DF-angle, so when the DF-angle changes, the MLA does not change immediately as well but takes a couple milliseconds to respond. This is however the first time the authors could show that the windlass mechanism is working during walking. As for RF-motion, the same pattern can be seen as with the MLA-angle; smaller DF-angles yield smaller RF-angles and vice versa (see Figures 1, 2, & 3). When the DF-angle becomes small enough, the RF-angle can even become negative, which in this study we labelled as a supinated rearfoot, whereas positive rearfoot values were labelled as a pronated rearfoot. Since the plantar fascia has a more medially located origin on the calcaneal tubercle, and as Caravaggi, Pataky, Goulermas, Savage, and Crompton (2009) found in their study that the highest can be seen near the medial calcaneal tubercle stress when elongating the plantar fascia, this would be in line with the expected patterns of foot motion in walking, meaning more stress on the plantar fascia means a more supinated rearfoot.

The data in this study would also suggest that when a dorsiflexion of the hallux occurs, the MLA and RF-angles become smaller, meaning there is an arch-height increase and a supination of the rearfoot. This corresponds with previous data regarding windlass functioning (Kappel-

Bargas et al., 1999; Nakamura & Kakurai, 2003; Caravaggi et al., 2009). These studies did not however investigate the difference between certain moments in the stance phase.

A question that arose when creating this protocol was whether treadmill walking was different from normal walking, and whether either would have a different effect on windlass functioning. First, when looking at the results of the normal walking tests, the same trend could be seen as with treadmill walking, being that the IC, MS and TO phases were all significantly different from one another for both the MLA and DF-angles. When comparing the two tests, some differences could be observed. For the DF-angle, the IC-phase and TO-phase were significantly different. This means that in treadmill walking, more dorsiflexion occurs at initial contact and at toe-off, so there seems to be some difference in the style of walking between normal and treadmill walking. However, since in the treadmill tests evaluation of the angles was performed manually, and in the normal surface walking test semi-automatic tracking was used to determine angles, no definitive conclusion can be drawn that treadmill walking has a different effect on the DF-angle. Despite the difference in DF-angle, these significant differences had no effect on the MLA-angle. This would suggest that when the goal of an assessment is to look at the MLA-angle, both treadmill walking and normal walking can be used, but when looking at DF-angles, clinicians should consider which is more appropriate for the situation of their patient and base their choice on that.

The data from the static tests showed that there is no difference between the pre- and post-tests for both the weight bearing and non-weight bearing situations, for both the MLA and DF-angles. There was however a difference between weight bearing and non-weight bearing for MLA-angle, but not for DF-angle. The examiners do note that bringing the hallux into passive dorsiflexion in a weight bearing position was much more difficult than in the non-weight bearing position. Despite this, total ROM for DF seems not to change. It is hypothesised that bodyweight (BW) influences MLA-motion during walking. Caravaggi et al. (2009) investigated the amount of tension through the plantar fascia during the stance phase expressed as a percentage of BW, but could not differentiate between lighter and heavier individuals given their small sample size of 3-three subjects. In a later study, Caravaggi et al. (2010) investigated the effect of walking speed on MLA-collapse and found that in higher speeds, a greater collapse occurred. Since higher walking speeds are associated with higher ground reactive forces, as a higher BW would also be, it would be interesting to investigate whether a higher BW also correlates with higher tension in the plantar fascia, and then test whether this higher tension results in a different MLA and/or RF-pattern during walking.

When then comparing normal walking to the static measurements, a big difference in MLA-angle can be observed. The magnitude of ROM in normal walking is about twice that of the static tests (see figure 8). In contrast, the magnitude of DF ROM was smaller in normal walking compared to static testing (see figure 9), with an average mean difference of 17.03°. This would suggest that MLA-rise is not just due to the windlass mechanism by itself, but also the effect of other factors. These other factors are most likely active mechanisms in the foot and ankle, as suggested by Tansey and Briggs (2001), who concluded from their study that the windlass mechanism accounts for approximately 50% of heel supination while the other 50% could be attributed to muscular components. While Thordarson et al. (1997) investigated the effect of the plantar fascia on MLA-height in cadavers, no info on that effect has been provided in vivo. Perhaps from these results it could be inferred that the windlass mechanism has approximately the same amount of effect on the MLA-rise in vivo.

The difference in MLA-angle between WB and NWB might suggest that tests such as the windlass test, where the toe is passively brought to dorsiflexion in a non-weight bearing position, are not suitable to assess dynamic function, and that weight bearing tests should rather be used to relate function to walking. However, given these results that show that normal walking yields about twice as much MLA ROM than static testing, neither the weight bearing nor the non-weight bearing tests should be used to assess dynamic windlass function.

When looking at the big picture of some results, it might be suggested that until more conclusive evidence is presented or normative values are established, it would be better to look at the pattern of MLA-rise and fall, instead of raw degrees. When assessing windlass function, it might be easier for clinicians to simply assess whether there is arch-height increase and decrease and RF pronation and supination during walking, rather than measuring the degrees and comparing them to other tests. It might not matter as much how many degrees are measured, but more so that these patterns can provide stability and propulsion during the stance phase.

Lucas and Cornwall (2017) concluded from their study that more research is necessary to investigate the dynamic relationship between MTP1 extension and changes in MLA-angle. This was partly delivered in this master's thesis. As seen in figure 10 (see Appendix F for figure), we can see a relation between MTP1 DF and MLA-angle changes in healthy individuals, as we excluded conditions such as hallux limitus and hallux valgus. When DF-angles become larger (meaning less MTP1 DF), the MLA increases as well. What's interesting to see is a confirmation of the pre-loading hypothesis suggested and investigated by Caravaggi et al. (2010). During the first moments of the stance phase (0-5%), there is dorsiflexion of the hallux which creates tension in the plantar fascia. It is seen in our data that this in turn leads to a sharper MLA-angle. This confirms the pre-loading hypothesis that states there is a pre-tensioning of the plantar fascia during early stance phase to provide more stability during the weight-acceptance phases (Caravaggi et al., 2010). During this weight-acceptance phase, a slight but steady increase in both DF- and MLA-angles can be seen, starting at approximately 10% of stance phase and ending at around 70%. This is around the point of the toe-rocker, where MTP DF starts to increase (smaller angles). As seen in the graph, the MLA does not react immediately but appears to have some lag. Therefore, the smallest DF-angle can be seen around 95% of stance phase, whereas the smallest MLA angle is seen at the very end of the stance phase. It would be expected that with the small increase in DF-angle during the last moments of stance phase (terminal stance), an increase in MLA-angle would also be observed. However, this is the moment the hallux acts as a rigid lever to provide propulsion of the foot. It could be hypothesized that to provide this function, the MLA-angle keeps decreasing (sharpens) until the foot loses contact with the ground. Caravaggi et al. (2009) showed however that the greatest tension in the plantar fascia can be observed around 80% of the stance phase, which is when the MLA-angle is the largest. Perhaps this late sharpening of the MLA is more so the effect of muscular components, and less so the effect of the windlass mechanism. The low value for the Spearman's Rho could also be explained by this graph. As said before, it seems the MLA lags behind the DF-angle when it changes, which makes their relation not very monotonic. The DF-angle could be increasing again while the MLA-angle still decreases. This could be the explanation why the Spearman's Rho is low.

More data is needed in different populations to establish normative values or ranges for windlass functioning in healthy individuals as well as certain pathologies (plantar fasciitis, hallux valgus/limitus,…). These graphs could then be compared to investigate different patterns and be linked to other pathologies of the lower limb such as achilles or patellar tendinopathy. While studying the effect of the windlass mechanism, Manfredi-Márquez et al. (2018) noticed movement throughout the entire lower limb when dorsiflexing the toes, which suggests that previous theories regarding this relation could be correct. Fuller (2000) provides an excellent biomechanical explanation for windlass involvement in the lower limb, while Prior (1999) hypothesised a relation between windlass dysfunction and several specific lower limb pathologies, such as plantar fasciitis, achilles tendinopathy, anterior knee pain, and even as far as low back pain. Unfortunately, the data of Manfredi-Márquez could not be validated due to flaws in the measurement system they used. They used large sensors which were attached to the foot at anatomical landmarks and connected to a system via Bluetooth which could interpret positional changes of the sensors reciprocally. The sensors were however quite large and heavy so as a result, there was a lot of motion of the skin relative to the bony segments when dorsiflexing the toes. Therefore, the objective data should be interpreted with caution. This was however a very interesting study because it was the first to test the kinematic involvement of the windlass mechanism in the lower limb. If more studies were conducted of this nature, provided they use a more refined measurement system, a link to lower limb pathology could finally be in the making. It has long been the belief of orthopaedic physicians and surgeons that the windlass mechanism is involved in lower limb pathology. It is our strong recommendation that future research regarding the windlass mechanism focuses on quantifying the dynamic relation between MTP1 extension and MLA-angle changes, as well as quantifying the kinematic relationship between windlass mechanism and the lower limb.

Lastly, ICC-values were calculated. The methods used in this study seems to be moderately reliable for intra-rater reliability in the treadmill test. When comparing the pre- and post-test for each rater, some statistical differences could be seen. For the RF-angles, this can be due to the fact that the assessment was based on a purely visual evaluation of rearfoot position. No markers were placed since previous studies did not specify adequately how they were placed, and therefore their methods were not completely reproducible. Furthermore, the authors wanted to test whether visual assessment was adequate to evaluate rearfoot motion. Given the ICC-values, visual assessment seems to be moderately reliable for intra-rater reliability, but poor for inter-rater reliability. The argument can be made for both RF and DF results, that the mean differences are not actually clinically relevant. For example, in the DF results, a difference of 1.2662° was found to be significant. Knowing that the ROM of toe dorsiflexion is about 40° in normal walking, as concluded from our own results of the 'normal surface walking-test', a difference of 1.26° does not seem relevant in a clinical assessment. The largest mean difference which was found to be statistically significant for the DF-angle was 2.05° (TO rater 2). The same argument can be made for certain RF-angles, where for the MS of rater 2 a mean difference of 0.34231° was found significant, in a total ROM of about 10°.

When looking at the inter-rater differences for the treadmill test, the same pattern was observed. Again, some mean differences were so small that we highly doubt the clinical significance of these differences. Except for the RF-angles, where some inter-rater differences of 3° or more could be observed, almost all differences were under 2° for DF-angles, and the highest MLA-difference was 1.06°. Even though these results for MLA and DF-angles were statistically different, the authors argue that they are not relevant in a clinical evaluation. The differences in RF-angle were however of such magnitude that they could be clinically significant. As mentioned earlier, rearfoot motion was more difficult to assess since no lines or markers were placed on the posterior side of the subjects' legs, but rather determined the RF-angle based on visual assessment.

Despite the significant inter-rater differences mentioned above, the ICC-values calculated from the data in this study suggest that the methods used have excellent inter-rater reliability for measuring MLA-angles for the treadmill tests, whereas the intra-rater reliability was good to moderate. The ICC-values for the normal walking test were good to moderate. For DF-angles these two tests showed good to moderate reliability. For RF-angles (treadmill only), intra-rater reliability was moderate, whereas inter-rater reliability was poor. This could be expected given the large statistical differences seen in the data-analysis. The poor values can once again be attributed to the method used to assess RF-motion.

ICC-values for the static tests were poor for MLA-angle and excellent for DF-angle. The method used for these tests was more difficult than anticipated. Semi-automatic tracking was also used to collect data for the static tests, however tracking did not seem to work accurately.

Therefore, constant manual adjustment of the angles was necessary. This is the most probable explanation as to why the ICC-values were low. It is unclear though, why DF-values were excellent and MLA-values poor, and not vice versa. Since the DF-angle needed constant adjustment whereas the MLA remained fairly stable with a small ROM throughout the whole test, it would be expected that ICC-values for MLA would be higher and DF-values lower.

A weakness of this study can be the use of everyday cameras (iPad). A reason for this is that during the treadmill test, images were sometimes blurry, since the hallux was the fastest moving part of the foot. This could have led to slightly different placement of the angles when measuring. In the normal walking tests where semi-automatic tracking was used, the same could have happened where the objects to be tracked (i.e. the markers) became blurry in the image and rendered the tracking less accurate. This would have happened mostly during loading response of the stance phase. A solution might be to use professional high-speed cameras, instead of the iPads used in this study which can record up to 240fps. The goal of this study however was to determine reliability of clinically feasible methods so high-speed cameras would be ruled out.

Another limitation of this study was that certain significant differences could most likely be attributed to a different placement of the markers on the foot. Some differences between pre- and post-tests could be seen in the same person, which suggests the markers were placed slightly differently on separate days. This would also be a factor when considering the reliability of the proposed method. However, with careful palpation and placement, this bias can be avoided.

7. Conclusion

Inter-rater ICC-values for the treadmill test were good to excellent for MLA and DF-angles, but not for RF-angles. Intra-rater reliability was good. For the normal walking inter-rater reliability was moderate. However, given there were no statistical differences when analysing the raw data, we hypothesize that both tests can be used fairly reliably to assess MLA-changes during walking. To evaluate RF-motion, another method needs to be devised as this study shows that visual assessment of rearfoot-angles is not reliable. Although clear patterns of pronation and supination could be observed, quantifying the motions using this method is not adequate. A better way to assess function might be to only look at the pattern of angle changes during walking for RF motion as well as for MLA-angles rather than measuring.

This study however provides a first step in the development of normative values and standardised testing. Results of static tests make it clear that they should not be used to assess dynamic function, as the difference in ROM is too great.

Future research should be focused on the dynamic relation between MTP1 dorsiflexion and MLA-angle changes, and the kinematic relation between windlass mechanism and the lower limb.

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9. Appendices

9.1 Appendix A Questionnaire (Dutch)

Vragenlijst voor deelnemers: Onderzoek Windlass mechanisme

Naam:
Geboortedatum:

(Schrappen wat niet past)

1. Heeft u orthopedische ingrepen/operaties doorgemaakt ter hoogte van de onderste extremiteiten (voet, enkel, knie en heup)?
   Ja / Nee
   Indien ja, specificeer :

2. Heeft u een afwijking van de 'normale' voetstand? (bv. platvoeten, holvoeten, hallux valgus (scheefstand grote teen), ...)
   Ja / Nee
   Indien ja, specificeer:

3. Heeft u in de voorgaande jaren blessures gehad ter hoogte van de onderste extremiteiten (heup, knie en/of enkelletsels)?
   Ja / Nee
   Indien ja, specificeer + datum:

4. Heeft u breuken gehad ter hoogte van de onderste extremiteiten?
   Ja / Nee
   Indien ja, specificeer:

5. Heeft u af en toe last en/of pijn aan de voeten?
   Ja / Nee
   Indien ja, specificeer:

6. Heeft u diabetes?
   Ja / Nee

7. Heeft u een bepaalde neurologische aandoening? (bv MS, CVA, perifere neuropathie, sensibiliteitsstoornissen,..)
   Ja / Nee

8. Heeft u een bepaalde ziekte?
   Ja / Nee
   Indien ja, specificeer:

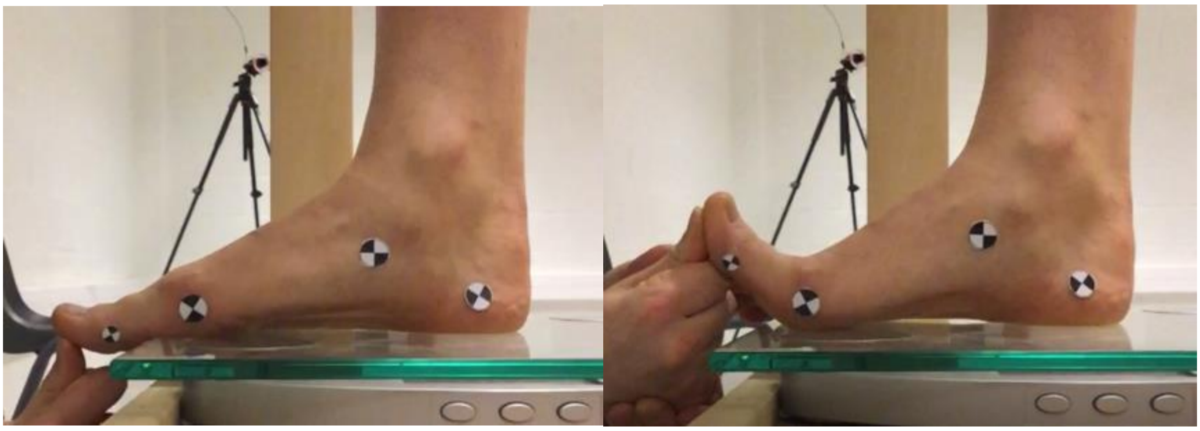
9. Heeft u gekende hartproblematiek?
   Ja / Nee
   Indien ja, specificeer:

10. Heeft u moeite met een tiental minuten aan één stuk te wandelen?
    Ja / nee

11. Welke sporten/hobby's beoefent u (hoe vaak / week)? Welk soort schoeisel draagt u hiervoor?

12. Heeft u voorkeur voor schoeisel dat u in het dagdagelijkse leven: Draagt u sneakers, geklede schoenen, hakken, orthopedische schoenen, ...? Draagt u steunzolen? (Hoe vaak/week)

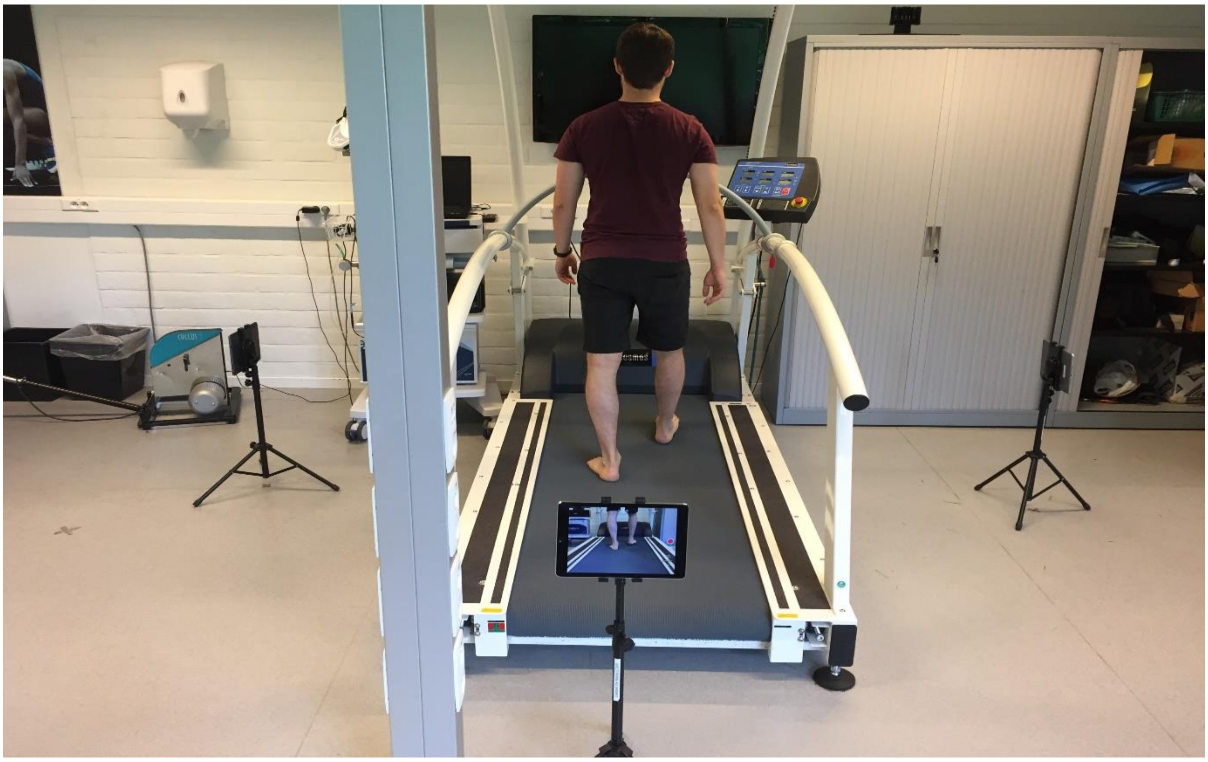
9.2 Appendix B Static testing & marker placement



As seen in the figure, manual passive dorsiflexion was used in the static tests. This was also the marker placement used in every other test.

9.3 Appendix C Testing positions & procedures

9.3.1 Treadmill tests

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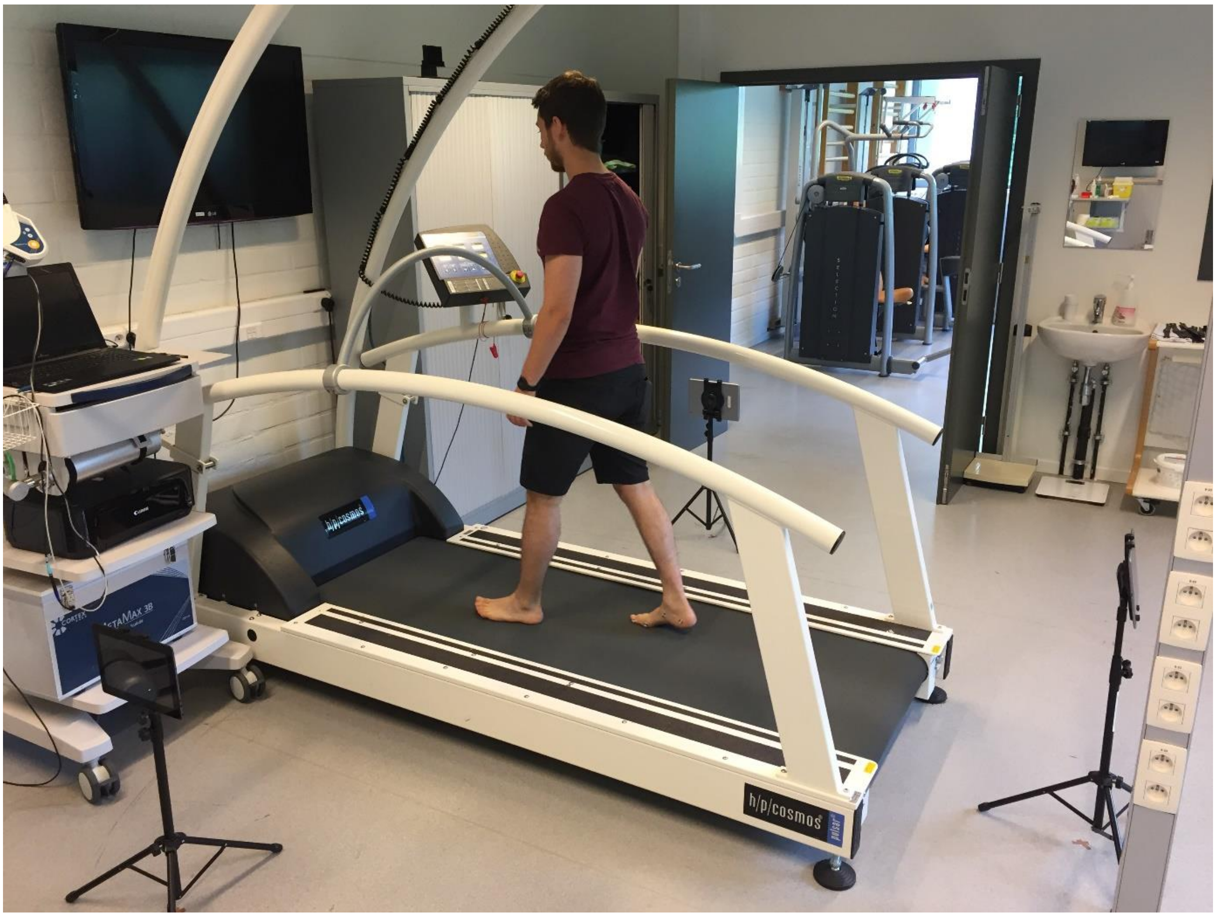
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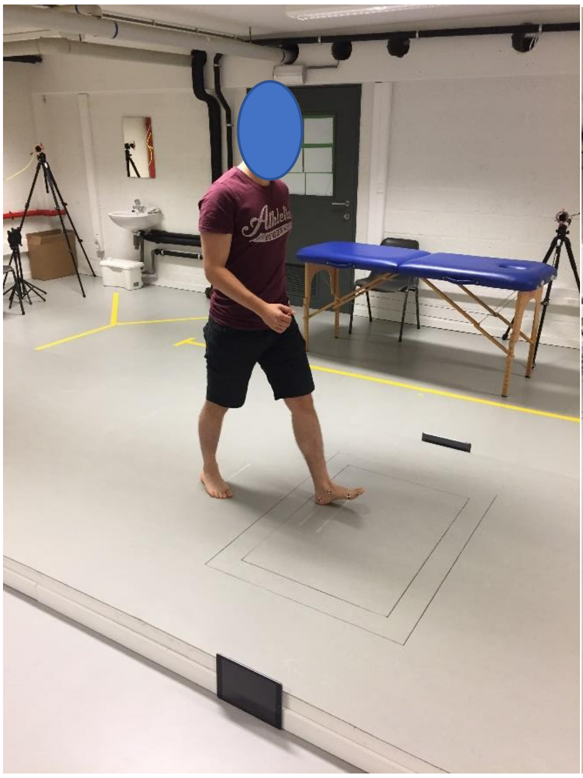
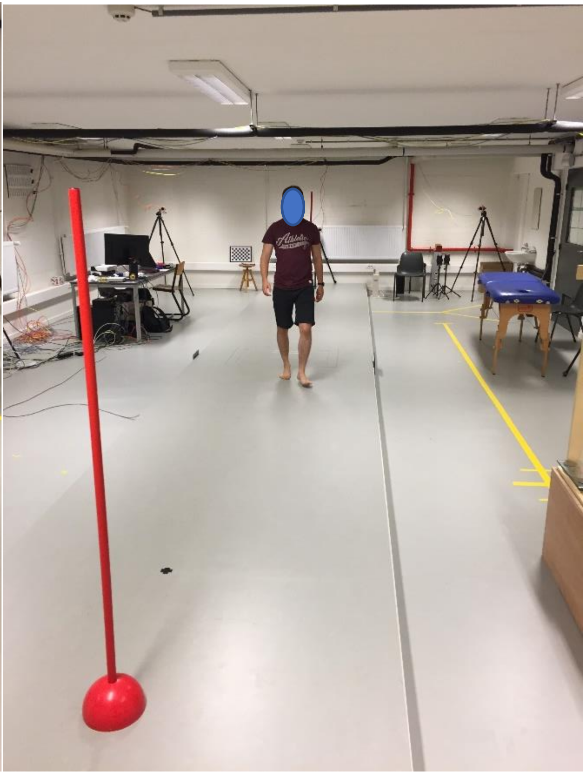
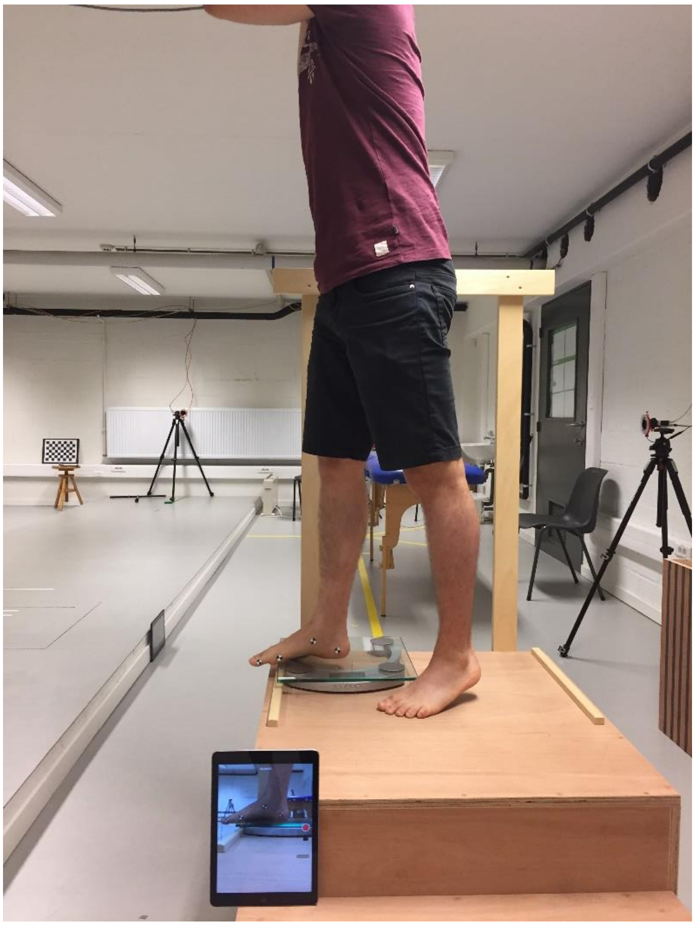
9.3.2 Normal surface walking





9.3.3 Weight bearing and non-weight bearing test



9.4 Appendix D ICC-values

ICC	INTRARATER TM	INTERRATER TM PRE	INTERRATER TM POST	INTERRATER NW	INTERRATER STATIC WB	INTERRATER STATIC NWM
IC MLA	0,76965	0,8974	0,9395	0,5658	0,3614*	0,183*
MS MLA	0,7648	0,9059	0,931	0,6186	/	/
TO MLA	0,77925	0,9018	0,9145	0,5879	/	/
AVG MLA	0,77123	0,90170	0,92833	0,59077	/	/
IC DF	0,6662	0,7709	0,6909	0,5957	0,8198*	0,8382*
MS DF	0,6456	0,7729	0,5686	0,5906	/	/
TO DF	0,7308	0,8055	0,7796	0,5146	/	/
AVG DF	0,68087	0,78310	0,67970	0,56697	/	/
IC RF	0,6792	0,39	0,3744	/	/	/
MS RF	0,50535	0,3319	0,2694	/	/	/
TO RF	0,6341	0,5125	0,4876	/	/	/
AVG RF	0,606217	0,411467	0,377133	/	/	/

*: For the static tests no values for IC, MS or TO could be calculated, hence these values represent the MLA-angles and DF-angles as a whole.

9.5 Appendix E Table of means of treadmill and normal surface walking data

Table 1

Means of treadmill and normal surface walking data.

MEANS (°)	Treadmill	Normal Surface Walking
MLA IC (SD)	126.523 (7.773)	127.126 (9.159)
MLA MS (SD)	132.230 (8.231)	130.868 (11.082)
MLA TO (SD)	121.904 (8.382)	122.163 (9.798)
DF IC (SD)	147.427 (8.091)	152.586 (8.449)
DF MS (SD)	166.355 (4.463)	165.369 (6.247)
DF TO (SD)	124.797 (7.695)	126.903 (7.625)

9.6 Appendix F Graph showing MLA- and DF-changes throughout the stance phase

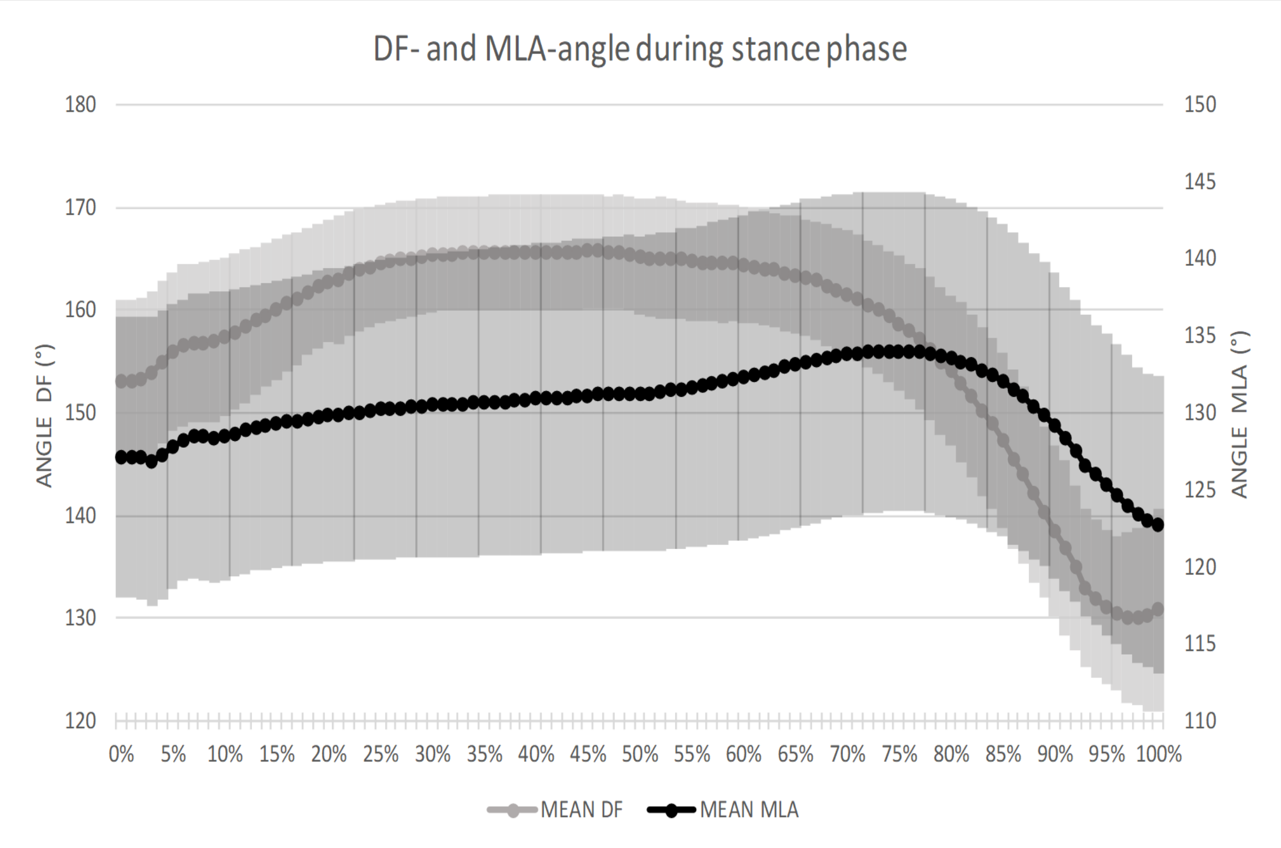


Figure 10: Changes of MLA- and DF-angle (with Standard Deviation-band) during the stance phase of normal walking. The curves were plotted using the mean values of all subjects of the normal surface walking test.

VOORTGANGSFORMULIER WETENSCHAPPELIJKE STAGE DEEL 2

DATUM	INHOUD OVERLEG	HANDTEKENINGEN
3/10/17	Voorbereiding metingen	Promotor: Copromotor: Student(e): Student(e):
18/10/17	Uittesten protocol + bespreking ethische commissie	Promotor: Copromotor: Student(e): Student(e):
6/12/17	protocol	Promotor: Copromotor: Student(e): Student(e):
24/01/18	Testmetingen + Dartfish test	Promotor: Copromotor: Student(e): Student(e):
3/05/18	opstart statistiek	Promotor: Copromotor: Student(e): Student(e):
		Promotor: Copromotor: Student(e): Student(e):
		Promotor: Copromotor: Student(e): Student(e):
		Promotor: Copromotor: Student(e): Student(e):
		Promotor: Copromotor: Student(e): Student(e):
		Promotor: Copromotor: Student(e): Student(e):

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