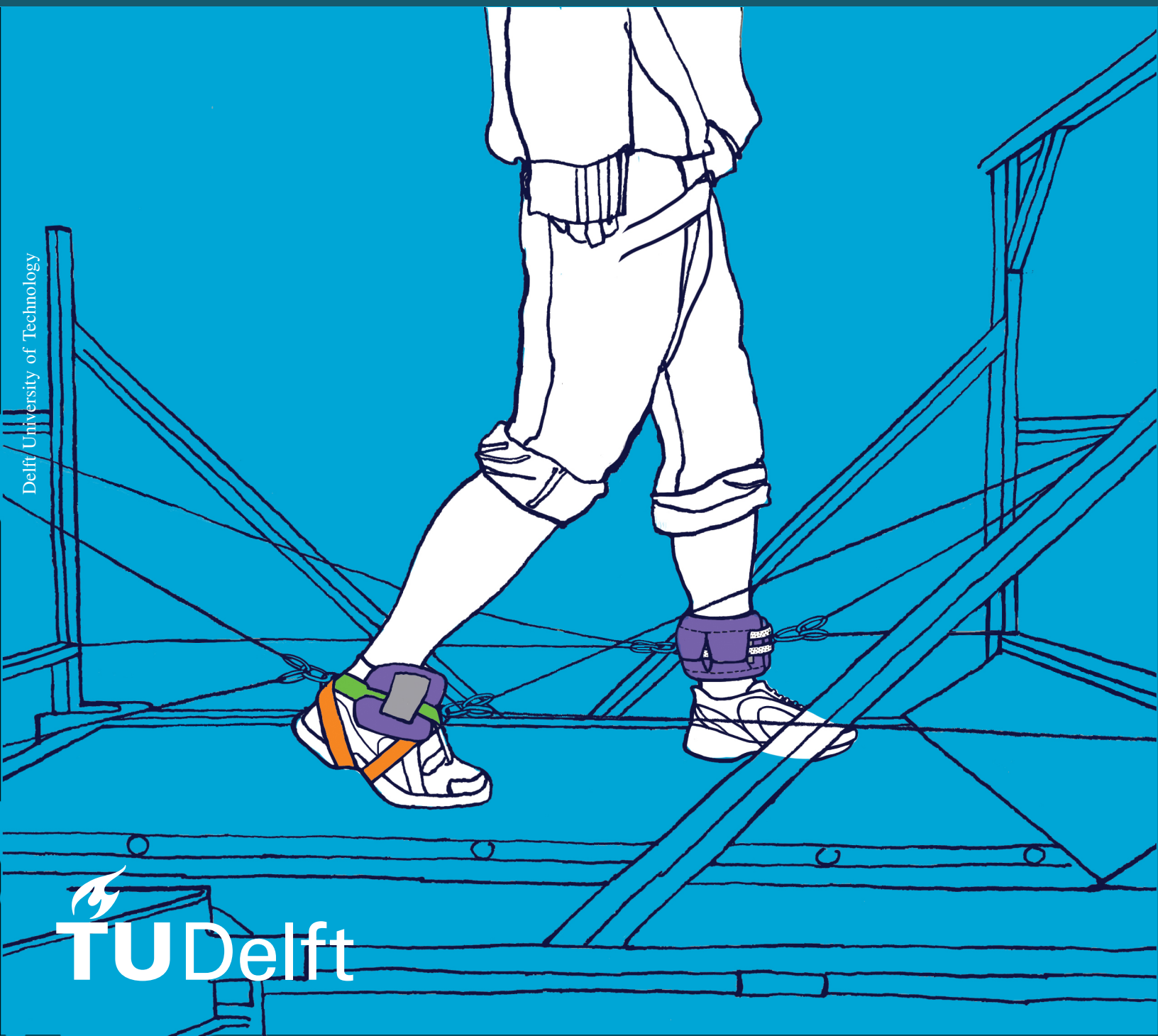
Design and evaluation of the connecting element between a human and the MATE, a tendon-based rehabilitation device for the lower body

MSc Thesis

Belle Cato Hopmans



Delft University of Technology

TU Delft

Design and evaluation of the connecting element between a human and the MATE, a tendon-based rehabilitation device for the lower body

MSc Thesis

by

Belle Cato Hopmans

to obtain the degree of Master of Science
at the Delft University of Technology,
to be defended publicly on July 16, 2024 at 09.30 AM.

Student number:	4362381	
Project duration:	March, 2023 – July, 2024	
Thesis committee:	Prof. Dr. Ing. Heike Vallery	TU Delft, supervisor
	Dr. Ing. Laura Marchal-Crespo	TU Delft, supervisor
	Dr. Katherine Poggensee	TU Delft, daily supervisor

An electronic version of this thesis is available at http://repository.tudelft.nl/.

TU Delft

ACKNOWLEDGMENTS

Dear reader, I'm writing my acknowledgements with a coffee in my hand on a bench in the sun, and when I think about my graduation project, I can only think… what a ride…! Lots of ups and downs taught me far more than just how to be and act like a bio-mechanical engineer, as this has been a journey of personal growth, too. I would like to use this part of my thesis to say thank you to everyone who helped me during this project. Firstly, I want to thank my supervisors, Heike, Laura and Katie. Dear Heike, thank you for your insightful feedback and infectious enthusiasm! Dear Laura, thank you for listening to all my updates, insightful feedback, and thoughtful questions. Also, thank you for the lovely chitchats and the compliments throughout the project. They helped me more than you might think and made me believe I was doing a good job:) Dear Katie, THANK YOU for everything! Thank you for being so patient, for listening, for feedback, for the time you spent on meetings with me, and thank you for the extra strolls when I needed to unwind after a meeting or a bad day. Thank you for your guidance and the uplifting words. You made such a difference during this long project!

Besides my supervisors, I want to thank Mark from the stroke rehabilitation centre for taking time out of his day to inform me about stroke rehab and test the MATE. Thank you, Mark from IWS and Juriaan from the welding department, for helping me realise my designs. Thank you to my mom and sister Tessel for helping me sew the designs during the late-night hours. Thank you to all the participants in my study for your time, honesty, and input. Thank you to Thomas and everyone at the MLN and Robot lab for your time and input when I asked for help. And thank you to my aunt Helen for all the giggles and the amazing front page you made!!!

Finally, I want to thank all my friends and family, not only during my graduation project but also during my master's, for all the kind and encouraging words! The hugs! The distractions! For listening! And all the laughs! You have truly made this a lighter experience! And although it is definitely not a standard thing to do, I want to thank me! Doing this project together with a second time ACL rehab was tough, but I'm very proud of how I handled the last year and how much I learned! So, thank you to myself for giving it my all and pushing through!

Belle Hopmans
Amsterdam, July 2024

CONTENTS

Acknowledgments i

Thesis Paper ii

I Introduction 1

II Methods 2
II-A Design . . . 2
II-B Experimental validation . . . 4

III Results 6
III-A Deliver *(second half)* . . . 6
III-B Experimental results . . . 7
III-C Design assessment . . . 10
III-D Case studies . . . 10

IV Discussion 11
IV-A The attachment designs are an extension of the MATE . . . 11
IV-B Compatibility with prescribed AFO . . . 13
IV-C User friendliness . . . 13
IV-D Hygiene of the designs . . . 13
IV-E Suitability for patients in the early stages of rehabilitation . . . 13
IV-F Use without supervision . . . 14
IV-G Blended care . . . 14
IV-H Limitations . . . 14
IV-I Future work . . . 14

V Conclusion 15

References 15

Appendix A: Double Diamond method: extra material 17
A-A Discover . . . 17
A-B Define . . . 19
A-C Develop . . . 22
A-D Deliver . . . 28

Appendix B: Force Calculations on Attachment Designs 32
B-A Force calculations & Velcro strap length . . . 32
B-B Comments after data analysis experiment . . . 32

Appendix C: Case Studies 33
C-A Case Study One . . . 33
C-B Case Study Two . . . 35

Appendix D: Experimental Results 39
D-A Additional Figures and Tables . . . 39
D-B Questionnaire data . . . 44
D-C Checklist data . . . 46

Appendix E: Design assessment 51
E-A Interview UNIP . . . 51
E-B Interview Create4Care . . . 52

Appendix F: Experimental Set-up 54
F-A MATE . . . 54
F-B Attachment Designs & test set-up . . . 54
F-C Force sensors . . . 55
F-D ZED mini camera . . . 55
F-E Questionnaire forms . . . 56
F-F Checklist . . . 56

Appendix G: Informed Consent Form 64

Appendix H: Python script used for experiments 66

Appendix I: Detailed force calculations document 74

Design and evaluation of the connecting element between a human and the MATE, a tendon-based rehabilitation device for the lower body

Belle Cato Hopmans
Cognitive Robotics Dept., Faculty of ME, Delft University of Technology

***Abstract*—Rehabilitation following a stroke plays a crucial role in functional recovery and the regaining of independence. Advanced rehabilitation robots such as the Lokomat, C-mill, Rysen, and ZeroG have become integral tools for gait rehabilitation. The MATE (Minimally Actuated Tendon-based Exercise Environment) is a novel rehabilitation device that could offer a wider range of motion compared to the Lokomat while providing support for leg movement, unlike the C-Mill, Rysen, and ZeroG. However, initial testing on the MATE indicated that this passive tendon-based rehabilitation device lacks a proper connection to the wearer. Consequently, this study was conducted to design and evaluate an attachment model, ensuring that the pulling forces are comfortably transferred to the body and that the MATE fulfils its purpose of supporting and enhancing the natural gait. The study compared two user-centred attachment designs based on user experience, comfort, usability, acceptance, ability to follow the natural gait, and the difference in pulling forces on the legs. The findings revealed that the bandage design exerted greater control over the legs, while the cuff design provided more guidance. Additionally, it was observed that adhering to the user's natural gait greatly influenced comfort and that walking with the MATE became easier with practice. Although the bandage design was preferred, the results suggest further iterations to enhance gait support, comfort, and usability.**

***Index Terms*—Robotic gait rehabilitation, Human-robot interaction, Human factor experiment, Comfort, User experience, Usability, Acceptance**

I. Introduction

AS technology continues to merge with the medical field, there is a growing focus on exploring the potential for robots to aid rehabilitation [1][2]. Annually, approximately 40,000 individuals in the Netherlands suffer from stroke, resulting in physical and cognitive impairments, such as loss of motor function and neglect of specific body parts [3][4]. Most stroke patients experience hemiplegia, where only one side of their body is affected. Rehabilitation is vital for restoring motor function, with the sub-acute phase—occurring from a week after the stroke up to six months later—being the prime time for rehabilitation [5]. Technological advancements have introduced robots such as the Lokomat, C-mill, ZeroG, and Rysen for gait rehabilitation, reducing the physical strain on physiotherapists [6][7][8][9]. However, the cost of these robots can amount to 300,000 euros, and specialised training may be required for physiotherapists to operate them. Additionally, when the robot necessitates multiple alignments and attachments to be secured to the user, such as with the Lokomat, the set-up can take valuable time out of physio training sessions.

Although these devices effectively aid in gait rehabilitation by reducing weight on the legs, ensuring exercise safety, having an unrestricted range of motion with the Rysen, utilising virtual reality applications with the C-mill, and, in the case of the Lokomat, moving the patient's legs for them, there are also drawbacks to these training methods. The C-mill, ZeroG and Rysen require the patient to move their legs themselves, and with the Lokomat, the legs are restricted to the robot's movements. A middle ground is found in the MATE, a tendon-based rehabilitation device [10][11]. This Minimally Actuated Tendon-based Exercise Environment (MATE) helps with gait rehabilitation by pulling the affected leg forward with the unaffected leg via elastic cables while wearing a body weight support system for safety. This gives the patient more freedom of movement during gait rehabilitation with the MATE than with the Lokomat and more support with walking than the C-mill, Rysen and ZeroG. Additionally, the MATE is more affordable than the aforementioned devices and does not contain any actuators, which makes it a passive and transparent rehabilitation device. Also, the MATE design's simplicity allows for a quick set-up, allowing more time to be spent on training during a physiotherapy session.

The previous study on the MATE developed the first physical prototype to examine the minimal forces exerted on the user during a healthy gait cycle [11]. The study stated that the wearer's attachment to the elastic cables was functional for the experimental purpose but lacked long-term comfort and ergonomics. During the human experiment in the study, it became evident that the connection needed improvement as it was reported to be uncomfortable and painful. The same study presented a new attachment between the wearer and the MATE but did not test it. However, this design was heavy, rigid, small, and non-adjustable.

Literature reveals that physical interaction between humans and robots is important for comfort and efficiency [12]. A study about soft exosuits for gait assistance states that comfort is essential for effectiveness and that, especially when the suit is worn for a longer period of time, comfort is crucial [13]. So, in the present study, a new connecting design was created and evaluated to contribute to the MATE's purpose of supporting and improving the natural gait during gait rehabilitation and enhancing the walking experience with the MATE. The design process followed the double diamond method, focusing on creating user-centred designs [14]. Two final designs were tested and evaluated, and the preferred model was recommended for further development.

II. METHODS

A. Design

The attachment design process is structured using the Double Diamond method *(Discover, Define, Develop, and Deliver)* [15] [16]. This framework supports understanding the problems and needs of end users, thus enabling the creation of a user-centred design. Effectively addressing design problems can lead to the development of an efficient solution [14].

1) Discover: The aim of the discovery phase is to gather a clear understanding of the problem to be solved and the challenges, desires, and needs of users in gait rehabilitation. To accomplish this, we conducted online research on (post-)stroke rehabilitation, visited a stroke rehabilitation centre, observed rehabilitation sessions, and held an extensive interview with a stroke rehabilitation physiotherapist. Additionally, we obtained Lokomat training evaluations from the aforementioned rehabilitation centre, which assessed patients' rehabilitation progress when training with the Lokomat.

Current state of post-stroke gait rehabilitation, as viewed by clinicians and patients. Rehabilitation aims to help individuals recover, restore function, and cope with temporary or permanent limitations. A crucial goal is to regain independence. The interviewed physiotherapist emphasised the importance of focusing on the walking quantity before walking quality. In the first stages of rehabilitation, the physiotherapist believes that becoming independent is more important than the symmetry of the gait cycle. As soon as functional recovery occurs, more focus should be on symmetry and quality. The physiotherapist stated that to relearn to walk in this new situation, a patient needs to develop new walking strategies, which is and needs to be a trial-and-error feedback process. However, this must be done in a safe way so the patient cannot get hurt and feels secure enough to test their balance capabilities. The physiotherapist believes that robotic exoskeletons, such as the Lokomat, mentioned in the introduction, can provide this safe environment but usually take over the gait cycle completely. This can be beneficial when that type of support is needed but not when one wants to stimulate the trial-and-error feedback process. The patient needs a full range of motion and room for adaptation and compensation of the whole body (freedom of movement). The physiotherapist also commented on other technological solutions, e.g., the ZeroG and the Rysen, see the introduction, which provide more freedom of movement during training than the Lokomat. However, in this form of training, patients have to move their affected leg with no support, which can be hard [17].

Furthermore, the physiotherapist emphasised the need for a fast set-up, which is currently a problem with commercialised exoskeletons. This is due to the numerous links, adjustments, and alignments, as well as the robot's user interface. Special training is required for individuals to operate these devices. Another issue noted about using exoskeletons is the need for constant supervision to ensure safety and make adjustments. The importance of blended care was also stated in the interview, emphasising the need for technology and care to work together. It was suggested that more (robotic) rehabilitation devices should be developed that require less patient supervision. Additionally, blended care for online rehabilitation enables people to train from home. Finally, the rehabilitation robots must be accessible, less expensive, and provided with a wide range of different-sized cuffs or attachments [17].

Despite the previously mentioned issues with the Lokomat, patient evaluations of rehabilitation progress while using the device indicate that patients enjoy working with it and notice improvements in their gait function [18]. They indicate having more confidence during walking, improving the swing phase of the affected leg, and the ability to walk longer distances. Also, to a slightly lesser extent, improvement was reported in walking speed, putting weight on the affected leg and safety during walking [18]. After using the Lokomat, brushing and swelling occurred mainly in the groin and not at the cuffs [18].

Current struggles with the MATE device. In the original design, the hooks at the end of the MATE's elastic cables were fastened around the user's lower legs using a lashing strap. Each leg had four cables connected to the cuff/strap, with the high cables attached to the inside of the leg and the low cables attached to the outside [11]. However, it was noted during that experiment that these attachments caused shear stress and eventually cut into the skin, hindering comfortable and effective gait support. A visualisation of the MATE's cable setup is shown in Appendix F-A.

Core personas. After exploring the problems in the hardware of stroke rehabilitation devices, the needs and wishes from the users' point of view were studied. The attachment will be used by both the patient, who will wear it, and the physiotherapist, who will apply it. Designing the attachment will require aligning the specific functional priorities both users need.

Two personas were created to represent the 'patient' during this study; see figure 9 in Appendix A-A1. A persona is a fictional character based on user research to represent different user types. Creating personas helps designers understand user needs, experiences, behaviours, and goals [19]. Our personas, Susie and Paul, suffer from different disabilities and require different rehabilitation. For instance, they differ in the type of disability, use of ankle-foot orthosis (AFO) during training, and points of attention in their gait pattern. Initiation of the swing phase in the gait cycle is the most challenging action for the patient, according to the interviewed physiotherapist [17]. Since all stroke patients might use the MATE device, the design process must consider all disabilities, challenges, and needs. Further information that inspired the personas can be found in Appendix A-A1.

Based on the interview with the physiotherapist and observation of training sessions at the stroke rehabilitation centre, we outlined a stroke rehabilitation session with the physiotherapist, as visualised in Figure 10 in Appendix A-A2. This journey map allows us to understand the environment in which our designed product will be utilised and how design features may impact the training session. The extra steps needed when working with current robotic rehabilitation devices during

training are shown. The essence of having a quick set-up time in a 30-minute physio session is clear when looking at the journey map. It was also mentioned as the most crucial issue of the interviewed physiotherapist and needs improvement [17]. More elaboration on the figure and the challenges and wishes of the physiotherapist during a training session can be found in Appendix A-A2.

2) Define: After learning about the issues and opportunities in gait rehabilitation after a stroke, the major problems were identified, and the objectives of this study were defined. The main objective is to create a solution for the interface of the cable attachments of the MATE device to humans. Achieving this goal must also involve resolving the main problems found in current gait rehabilitation devices. Consequently, while designing the new attachments to connect the wearer to the elastic cables, extra attention must be given to enhancing the MATE's ability to support and improve the wearer's natural gait. This involves comfortably transferring pulling forces to the wearer while preserving full freedom of movement and without hindering adaptability. In addition, the user interface should be easy and attractive to use, and the setup process should be quick.

Furthermore, in addition to ensuring the aforementioned points in the design process, other considerations include hygiene, the use of walking aids, and blended care. A comprehensive list was created containing all the features to be considered in the design process, considering the challenges and preferences of both patient personas and the physiotherapist. From this list, a set of requirements was derived, outlining all the necessary points to achieve the main objective and a wish list featuring important points to include but not essential. All the design features, with specifications, are listed in Appendix A-B1, and both the list of requirements and the wish list are shown in figure 11.

3) Develop: Throughout the development phase, we considered various solutions for the problem outlined in the *Define* phase. Our selection process commenced with an examination of similar problems and solutions in different industries using a mind map (see Appendix A-C2), the identification of necessary design features to meet the specified requirements, and an interview with an orthopaedic device maker from a rehabilitation centre. Subsequently, the selection process involved creating sketches and concepts in SolidWorks, exploring different materials, performing force calculations, and assembling and iterating concept designs. Simultaneously, we explored options for connecting the elastic cables of the MATE to the attachment designs. Meanwhile, we consistently ensured that the design aligned with the outlined requirements.

Design features impact design performance. Before deciding on design features, the list of requirements was completed by examining how features affect design performance. The feature choices can affect the type of gait support, such as lifting the foot at the start of the swing phase. In addition, including an AFO during training can accommodate and fulfil desirable features, such as supporting ankle dorsiflexion during the swing phase. Another performance-influencing feature is the placement of the attachment with respect to how the forces are applied to the user during different gait phases. In the new design, the attachments were placed around the ankle, which is an effective location for applying for assistance. Additionally, the cables were bundled at the front and back of the design, aligning with the direction of the highest pulling force, optimising the force path and reducing shear stress.

In addition to exploring the impact of design features on the gait cycle, we also examined their influence on comfort. It is important to select features that contribute to light and compact designs to enhance comfort. Choosing design features that facilitate the transfer of pulling forces to the user in all intended directions enhances the synergistic interaction between the MATE and the user. Other design choices that affect comfort are uniform pressure distribution and good fit. Design simplicity is important for an easy user interface and a quick setup. A full elaboration of the design features influencing the points on the list of requirements is given in appendix A-C1, which also contains more information about design choices in terms of hygiene, and how the shape and materials of the design can contribute to a strong design.

Design concepts and iteration. Two final concepts were developed: a cuff design and a bandage design. Both designs are attached to the ankle. However, the cuff design is placed just above the ankle, exploiting the tapered shape of the leg to secure the attachment, while the bandage design is connected around the ankle joint with two straps—one wrapping around the ankle and the other crossing under the foot to create a sling shape. Both designs use a Velcro closure.

We tested the initial designs for functionality with the MATE by walking in both designs for one hour and pushing the designs by making large and abrupt movements. Although both attachments allowed comfortable walking during the whole hour, further development was needed for simplicity, hygiene, and usability. An interview with an orthopaedic device maker provided insight into the potential use of the concept designs with an AFO. This revealed that the type of shoe and heel height influence the function of the AFO and that the placement of an attachment around the shoe was preferred. The interviewee indicated that the pulling forces at the chosen connection points of the cables with the attachment were unlikely to affect the AFO function negatively, but this needed to be tested further.

We successfully tested the second round of prototypes with updated design features by repeating the same tests as in the first round. We noted the improvements to enhance the final version of the two designs. Appendix A-C4 provides more information about the iteration process.

The selection of materials for the designs was influenced not only by their properties but also by the list of requirements. Common materials used in the designs include cold foam padding, Velcro closures, synthetic leather, lashing straps, aluminium, and stainless steel (see Appendix A-C5). The materials are readily available and affordable.

During the development phase, force calculations were performed to analyse the force pulling on the designs in various directions. We assessed normal usage scenarios and worst-case

situations, such as falls or closure strap malfunctions. With the information we obtained, we calculated the necessary amount of Velcro for the closure Velcro straps to ensure safe utilisation of the attachment designs. More details on force calculations and safety measures are provided in Appendix B.

4) Deliver (first half): During the final phase of the design process, we developed, tested, and assessed two final attachment solutions. Through experiments involving unimpaired individuals, we gathered data on their experience with the MATE, their acceptance of the MATE with the designs, and the product's usability. Additionally, we collected their kinematic data and measured the pulling forces exerted on the participants' legs. After analyses, we made a choice for the most promising design solution.

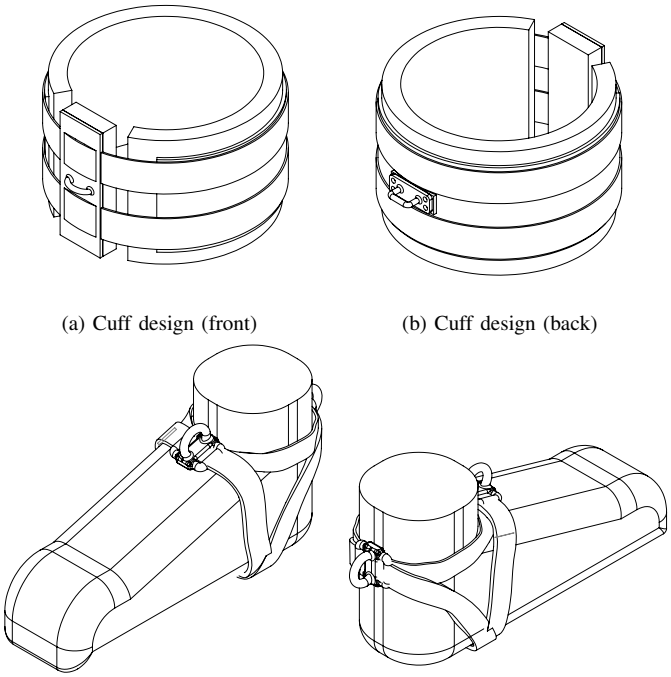
Final design concepts. The cuff design consists of a C-shaped aluminium back and a rectangular front, held together with Velcro straps. This design includes a thick layer of cold foam padding on the inside, making it fit snugly on the leg just above the ankle. We used synthetic leather to cover the padding, improving the overall hygiene of the design. The steel anterior and posterior hooks serve as connection points for the cables. Figures 1a and 1b display the cuff design without synthetic leather covers around the foam.

The bandage design comprises a harness-like structure around the ankle and foot, made of lashing straps with Velcro closures at the ends. Two stainless steel buckles customised for this design secure connection points to the straps. Pads of cold foam covered with synthetic leather are added between the ankle and the straps for additional comfort. Figures 1c and 1d display the design without the pads. In both designs, stainless steel carabiners fasten the elastic cables to the designated hooks or buckles.

Differences between both designs. Apart from the differences in the location of the attachments on the leg, the two designs have additional functional differences. The bandage design is compatible with any AFO as it is secured around the shoe, whereas the cuff design cannot accommodate an AFO unless it runs tightly along the body, due to its connection to the leg. Both designs come in different sizes, but the bandage design offers more size options due to the non-adjustable straps and the necessity of a proper fit. The (upper) strap around the ankle and the (lower) strap wrapping the foot are available in various lengths to ensure an optimal fit for the user. The cuff design is more adaptable for individuals of different sizes and has fewer size variations. Lastly, when not in use, the cuff maintains its shape through the aluminium base, while the bandage design can be flattened and folded.

B. Experimental validation

Eight unimpaired adults participated in the experiment to assess the impact of the two different attachment designs. The following five questions were asked to help differentiate between the designs and determine the most promising one.



(a) Cuff design (front) (b) Cuff design (back)

(c) Bandage design (front) (d) Bandage design (back)

Fig. 1: Concept of the final designs. A SolidWorks overview of the front and back of the cuff and bandage design.

Each sub-question provides specific data required for a part of the conclusion. The answers to the questions are expected to overlap, leading to new correlations and conclusions.

- Do the forces exerted on the human change within and between the different designs?
- Do the participants' kinematics change when walking with the MATE in the different designs compared to walking during the experiment's baseline?
- Are there differences in user experience between the two attachment designs?
- Does the acceptance of the MATE change between both designs?
- Are there differences in usability between both attachment designs?

Both designs were expected to comfortably transfer the pulling forces of the MATE to the wearer during the experiments. However, due to their distinct design properties, they were anticipated to excel in different aspects, thus impacting the gait cycle differently. The bandage design has the potential to exert more control on the legs and feet due to its harness-like structure around the feet, providing support in lifting and placing the feet in specific positions on the ground. This would be evident from the high pulling forces on the legs, similar or higher step height than the baseline step height, and anomalous foot positions compared to the participant's baseline gait. Additionally, changes in pulling force at the beginning and end of the 20-minute gait session would reveal the MATE's and the user's cohesiveness. The answers to the user experience and acceptance questionnaires, as well as notes and quotes from the participants, would reveal whether the MATE with the bandage model was helpful or opposed the

participant's natural gait. Additionally, they provide another point of view and more insights into the effects of high pulling forces on kinematics, comfort, and the type of support.

The cuff design was expected to have a more guiding effect on the gait pattern. Fitting around the ankle instead of the foot was anticipated to offer the wearer greater control over their feet. This was expected to manifest in lower pulling forces on the legs during the 20-minute walk and kinematics similar to the baseline. While the cuff design was assumed to be less influential on the participant's gait pattern than the bandage design, it was still expected to assist in the forward movement of the gait cycle. This would be evidenced by the similarity of step length to the baseline and the participant's comments indicating that the MATE is helpful without overly restrictive.

The comments would also provide insights into the comfort of the design, where the cuff was anticipated to be more comfortable around the ankles than the bandage model. However, comfort is influenced by various factors (e.g. fit, synergistic interaction with the device, and pressure distribution), and personal preferences may also play a role in determining the preferred design, which would be reflected in the participants' varied responses and preferences. Furthermore, it was presumed that the cuff design would be easier to use at first glance. However, once the workings of the bandage design are grasped, it may become the more user-friendly option. This would be clear through shorter participant donning and doffing time with the cuff design, and a higher usability score on the questionnaire would yield equivalent results. Additionally, a change in the experimenter's donning and doffing time would show what experience does to the usability of the designs.

The cuff design was expected to be the preferred choice in this experiment, as it is anticipated to be more usable and allow for kinematic movements similar to the baseline. However, the MATE aims to provide a level of support that could enhance a patient's gait pattern and walking quantity. Therefore, if the cuff design outscores the bandage design in this experiment, the bandage model should not be disregarded, as it is expected to offer different features and qualities and may fulfil different forms of gait assistance (e.g. lifting the foot during walking).

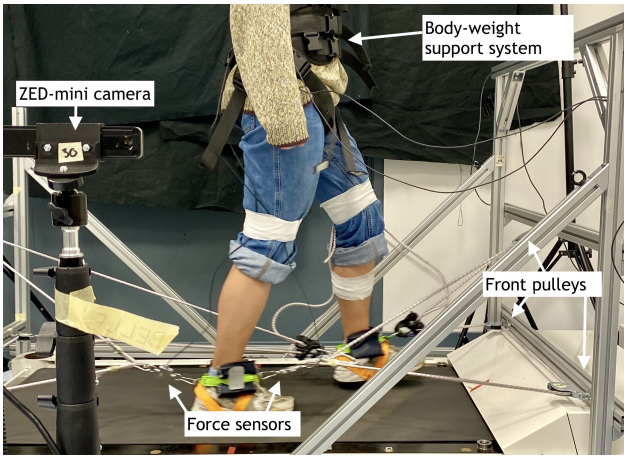
1) Participants: Eight unimpaired adults participated in the experiment to evaluate the impact of the two attachment designs. The group consisted of 4 men and 4 women, aged between 22 and 63. Additionally, their weights and heights were documented in the experiment's checklist (Appendix F-F). This research was approved by the Human Research Ethics Committee of Delft University of Technology. Before the experiment, all participants provided informed consent by reading and signing a form, which is available in Appendix G.

2) Experiment protocol: The experiment commenced with explaining the donning and doffing of the designs to the participant, and simultaneously, the correct size of the attachment designs was checked. Before walking with the MATE, a baseline of the participant's kinematics was established in 5 minutes. Subsequently, each participant walked for 20 minutes with the MATE per design, wherein kinematic data and forces on the right leg were obtained. The order in which the participant tries on the cuff and bandage design is randomised. Additionally, two questionnaires were utilised to glean insights into user experience, acceptance of the MATE with the designs and the acceptance of the designs alone. Further information was gathered during the experiment on the donning and doffing time by the experimenter to test improvement in experience. After the gait sessions, the participants donned and doffed the designs and subsequently completed a third questionnaire to gain insights into the usability of both designs. Comments and quotes from participants during the gait sessions with the MATE were recorded in the experiment's checklist, providing extra insights into their experience, comfort, and acceptance of the MATE. Finally, treadmill speed, design order of walking with the MATE, and demographic information (gender, age, height, weight) of each participant were also documented in the experiment's checklist.

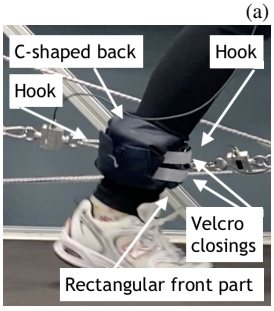
3) Experimental set-up: The MATE comprises a frame construction with four elastic cables and six pulleys. The same set-up was used as in the previous study on the MATE, and the same pulleys and elastic cables were used [11]. The frame was reassembled with the information from Kieft's Table 1 [11], as were the positions of the pulleys. Before a participant's first walking session with the MATE, the cables initially started at standard lengths but were adjusted using the cable tension hooks (shown in Appendix A-D3) if the participant found the cables to be exerting too much force on their legs, affecting their normal walking pattern. The same cable lengths were maintained for both walking sessions with the MATE for each participant. The test set-up is displayed in figure 2, where the position of the ZED-mini and force sensors are shown. More detailed information is given in Appendix F.

4) Safety measures: Before commencing the experiments, the attachment designs underwent testing for comfort and safety by walking with the MATE and both designs for an hour. To address potential risks related to shear stress and hygiene, synthetic leather material was utilised for the parts of the designs that touch the skin, offering easy cleanability and softness. Participants wear socks under the designs for added hygiene and to reduce shear stress. The body weight support (BWS) system of the Lokomat is employed during all gait sessions to prevent falls. Before the gait sessions started during the experiments, the participants tested the BWS. Furthermore, the participant and experimenter had access to their own emergency stop button to immediately stop the treadmill if necessary. Finally, during the test session, we attempted to step with one foot over the other, but this movement was impossible due to the tension in the cables at the standard length.

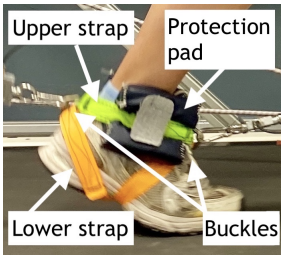
5) Biomechanics: Two load sensors are placed between the cable ends and the right attachment design. Each KDC 40S 1000N load sensor measures the pulling force caused by the MATE elastic cables on the right attachment designs' anterior or posterior hook and, consequently, on the front and back of the participant's right leg. The force offset was determined by a sensor calibration conducted at the beginning and end of each session. During each gait session with the MATE, the pretension on the anterior and posterior of the design was obtained when participants stood at a standstill in the middle of the treadmill, indicated by red tape on the sides of the treadmill. The force sensors on the attachments measured the



(a) Test set-up



(b) Cuff design



(c) Bandage design

Fig. 2: This figure visualises the test set-up during the experiments.

pulling force for 10 seconds before and after walking, and the averages were calculated for the front and back of the design.

A StereoLabs ZED mini Stereo Camera, a depth and motion-sensing camera, was used to capture kinematics. The onboard software can detect anatomical landmarks (i.e., neck, hip, knee, and ankle) as well as standard video. Real-time force readings are displayed in the experimenter's view (top left corner) of the ZED-mini video. The force sensors and kinematic data are both synchronised offline. For more details on the experimental set-up, refer to Appendix F.

6) Questionnaires: Three different questionnaires were employed to gather data on the participants' user experience during the gait sessions, their acceptance of the MATE with the designs, their acceptance of the designs themselves, and information regarding the design usability. We utilised a modified version of a User Experience Questionnaire (original [20]), the original Van Der Laan acceptance questionnaire [21], and the original System Usability Scale [22]. The scores were calculated using the scoring rubrics of the original questionnaires. Further details are provided in appendix F.

7) Donning and doffing time: The experimenter used a stopwatch on an iPhone 11 to time the participants' and her own donning and doffing. This included timing the process of putting on or taking off both designs (e.g. start the stopwatch, put on the first cuff, put on the second cuff, stop the stopwatch).

8) Cuff height: The cuff height was measured three times: initially right after donning, then when the cables were attached to the design but before the gait session, and finally immediately after the gait session with the cables still attached. This measurement was taken using a tape measure from the bottom of the cuff to the ground.

9) Notes and quotes participants: During the gait sessions, observations and quotes related to the MATE, experience, and comfort were recorded by hand. Only relevant comments for this study were translated into English.

10) Data processing: After the experiments were conducted, all the data was imported into Matlab R2017b for analysis. First, the data was cleaned. The kinematic data was already filtered in the ZED-mini camera, so the outliers were removed only from the force sensor data. Subsequently, each dataset was resampled with a frame rate of 32 Hz. The data was then segmented into time periods representing the entire sessions and periods of one minute. For analysis, minutes 2 and 4/19 of each gait session were chosen as they provided enough data to demonstrate potential differences in participants' gait patterns throughout a session. The first minute and last minute were excluded to focus on steady-state.

For each condition, we computed the average step length, average step height, maximum, minimum, and average forces on the front and back of the attachment design and obtained the ankle trajectories. We only utilised the kinematic data of the right ankle as the data detection confidence from the left ankle was noticeably lower, and more data points were missing. Moreover, the side-to-side position of the right ankle was omitted since the ZED-mini was not calibrated for depth (z-position) in the test setup before the experiments.

III. Results

A. Deliver (second half)

The final designs, when connected to the MATE, are depicted in figure 2b and 2c. Both designs are available in small, medium, and large sizes to ensure a proper fit for the wearer. The front is interchangeable with the cuff, and for the bandage design, the upper and lower straps can be adjusted for a perfect fit. Both designs incorporate foam padding for comfort, covered by synthetic leather for hygiene. Furthermore, both designs feature Velcro closures, where the hard side of the Velcro straps always faces away from the body. This ensures that if the Velcro closing sticks out and touches the user's skin, it is always the soft side of the Velcro. However, both models are designed in a way that the Velcro should not touch the skin.

1) Final cuff design: The cuff design (figure 2b) is lightweight, weighing 194 grams. Each cuff includes several replaceable parts, such as covers, closing straps, and front and back components. The cuff design utilises differently coloured Velcro to indicate which closing strap or flap corresponds with each other.

2) Final bandage design: The bandage design (figure 2c) is similarly lightweight, weighing only 107 grams. The separate upper strap, lower strap, and protection pads can be easily replaced as needed. Vibrant and different colours were chosen for the upper and lower straps to create clarity and help users identify the correct positioning of the straps when applying the bandage. Additionally, customised buckles have

been designed with matching strap colours to indicate the intended placements.

3) Connection MATE to attachment designs: In this experiment, the cables from the front of the MATE are gathered using a carabiner and attached to the front of the design on the right leg at the designated hook. The same process is repeated on the back of the design. Throughout the experiment, force sensors are connected between the right attachment and the cables. On the left leg, the cable ends are attached to a cable tension hook that allows for adjusting the length of the elastic cables. Once again, the tension hooks securely fastened the MATE cables using a carabiner, either to the front or back of the design. Appendix A-D provides more detailed information on the designs and the connection.

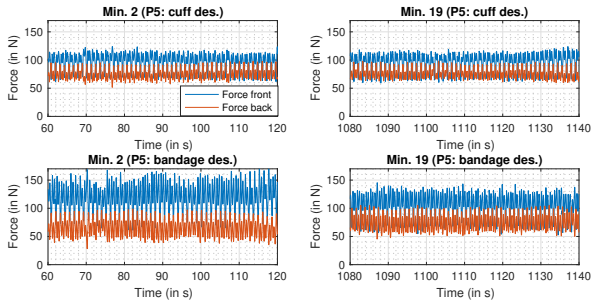


Fig. 3: Forces on the front and back of the attachment designs during the 2nd and 19th minute for Participant 5. The top row shows min. 2 and min. 19, respectively, of the cuff design. The bottom row shows the min. 2 and min. 19 of the bandage design. Data of all participants can be found in figure 26.

B. Experimental results

1) Pulling force differences between designs: The pretension in the cables during walking with the cuff design and the bandage design, along with the corresponding unstretched cable lengths, are shown in table X in Appendix D.

In table I, all the averages of the forces are listed, i.e., for the front and back force sensors, the average of the maximum and minimum forces during each walking session and the overall average pulling force during a walking session with the MATE. For all participants, the average pulling force on the design's anterior is higher when wearing the bandage design compared to the cuff design. Conversely, the average pulling force on the design's posterior was highest when using the cuff design for 7 out of 8 participants. Across all participants, the maximum force exerted on the front of the designs was 190 N. The corresponding force exerted on the back of the design at that time was 32 N. The maximum force measured in the sensor at the back of the design was 126 N, with a matching 52 N at the front of the design at that time. Both maxima were recorded while using the bandage design. Figure 25 in Appendix D shows the box plot with the average maximum force at the front and back of both designs.

An overview of all the forces during the 2nd and the 19th minute of each participant is shown in figure 26 in Appendix D; a representative participant is shown in figure 3. The force data from this participant serves as a prime example of the variations in exerted pulling force that can occur between walking in the cuff and bandage design, but also between the 2nd and 19th minute of the same design. The cuff and bandage designs differ in terms of the fluctuations between the average maximum and average minimum force peaks. In all participants, the difference between the maximum and minimum peaks is greater for the bandage design compared to the cuff for both the forces acting on the front and back. Furthermore, the forces exerted on the cuff design appear to remain similar between minutes 2 and 19, the forces in the bandage design seemed to decrease over the session time.

2) Relationship between forces and kinematics: Figure 4 depicts the forces acting on the right leg throughout various phases of the gait cycle. At the moment of the heel strike, the force at the back of the design is at its maximum, and the force at the front is at its lowest. During mid-stance, the forces shift as the force at the front increases and the force at the back decreases. At the moment of toe-off, the design's back force briefly increases, and the design's front force slightly decreases. However, immediately after, in the mid-swing, the forces return to the same level as during mid-stance, so the design's front force expands, and the back force drops slightly. This periodic shape of the forces is observed in both attachment designs.

3) Kinematic data: We evaluated changes in the leg kinematics between baseline (walking without the MATE) and walking with each one of the designs. In figure 5 the right ankle trajectories are displayed for each participant during the baseline condition superimposed onto the ankle trajectories when walking with each design at minute 19. The ankle trajectories with the designs seem to differ from the shape of the baseline ankle trajectory. In the following, we will evaluate the step length and height under the three conditions, namely baseline, cuff, and bandage designs.

The left box plot in figure 6 displays the average step lengths for each condition. In general, using the MATE in its current designs leads to shorter step lengths compared to the baseline. The average step heights are shown in the middle plot. In general, the step height is slightly reduced when walking with the cuff design, while the step height is more reduced with the bandage design. As the step length and step height data are obtained from the right ankle marker and not from the foot, an additional plot of the average step height is shown; see the right box plot in figure 6. Here, the ankle height of the participants in mid-stance (a snapshot, not average) is added to their average step height to get another perspective on the average step/ankle height. Again, the bandage design displays the lowest average step height. However, in this plot, the gap in height between the conditions is smaller. The median values of all plots are noted in table II.

4) Questionnaires: The questionnaire results are shown in the box plots in figure 7. The median values are reported in table III. Appendix D-B contains the box plots displaying the different participants' scores in figure 29, and tables reporting the questionnaire scores of each participant; tables XII and XIII.

The results from the **User Experience** questionnaire in-

TABLE I: Average forces on the front and the back of the right attachment design. The average of the maximum force peaks, minimum force, and the total data of the cuff and the bandage design are noted. An enlarged version of the table can be found in Appendix D in figure XI

Pulling Force (in N)	P1		P2		P3		P4		P5		P6		P7		P8	
	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design
Average max. force front	117.68	138.06	108.41	121.88	101.78	107.20	104.68	125.95	114.92	140.58	104.92	110.75	100.55	108.63	96.22	115.68
Average min. force front	81.06	81.30	65.22	67.72	81.18	80.17	59.43	60.79	63.73	65.43	59.49	53.50	57.08	55.36	63.10	69.88
Average pulling force front	**102.48**	**114.02**	**91.86**	**97.37**	**94.64**	**95.99**	**88.09**	**98.41**	**96.21**	**108.79**	**86.76**	**87.74**	**84.09**	**86.20**	**83.94**	**96.32**
Average max. force back	89.47	94.37	98.16	97.62	110.22	109.49	103.04	110.40	95.53	99.64	101.80	99.90	106.19	114.24	101.82	95.03
Average min. force back	65.23	59.93	69.24	59.57	81.18	67.48	75.45	68.06	63.51	54.31	70.19	58.27	73.03	70.33	76.68	58.34
Average pulling force back	**76.41**	**75.11**	**81.70**	**77.52**	**93.86**	**85.70**	**86.85**	**86.06**	**77.38**	**73.32**	**83.78**	**75.14**	**86.94**	**88.56**	**86.96**	**74.80**

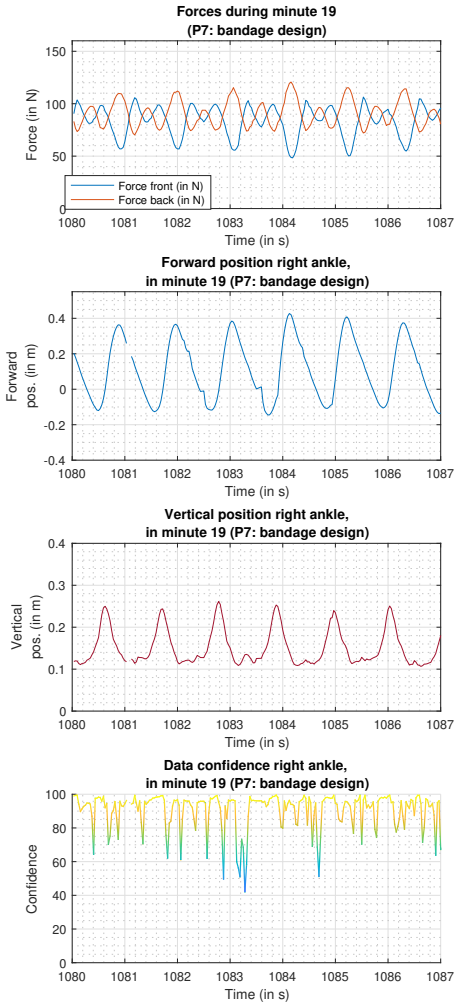


Fig. 4: The figure depicts the kinematic data and forces with the bandage design of Participant 7. It includes the forces at the front and back on the right ankle, as well as the forward and vertical motion of the ankle. The reliability of the kinematic data is presented in the bottom graph.

dicate that, on average, participants had a better experience with the cuff design in comparison to the bandage design. However, it is worth noting that not all participants favoured the cuff design over the bandage design. In particular, three out of eight participants preferred the bandage design. These

TABLE II: The median values of the average step length, average step height, and average step height + ankle height of the three different conditions are presented in the table.

Median value of...	Average step length (in m)	Average step height (in m)	Average step height + ankle height (in m)
Baseline	0,581	0,157	0,253
Cuff design	0,529	0,147	0,245
Bandage design	0,525	0,1	0,228

specific participants also had a longer front high cable than the standard length.

Overall, both designs scored high in **Usability**, especially the cuff design with a median value of 85 over 62.5. The cuff design received higher usability rankings than the bandage design, with six out of eight participants rating the cuff higher in this aspect.

Lastly, a questionnaire on **System Acceptance** was conducted. The acceptance of the designs was rated, as well as the acceptance of the MATE while wearing the designs. The questionnaire made a distinction between the usefulness scale and the satisfaction scale. The average scores are overall positive. Regarding the MATE acceptance, the cuff design received a higher average satisfaction rating (six out of eight participants ranked the cuff higher than the bandage). The bandage design scored higher than the cuff on average in terms of usefulness for MATE acceptance (six out of eight participants favoured the bandage over the cuff). In short, the median scores are close in acceptance between designs, with the cuff scoring highest on both the usefulness and satisfaction scales.

TABLE III: This table reports the median values of the questionnaire box plots in figure 7. The U is short for Usefulness and S for Satisfaction. The questionnaires were scaled from -2 to 2 for user experience and system acceptance and 0-100 for design usability.

Median value of...	User experience	Design usability	MATE acceptance		Design acceptance	
			U	S	U	S
Cuff design	0,8	85	0,8	1,0	1,3	0,6
Bandage design	0,6	62,5	0,9	0,6	1,0	0,4

5) Checklist data: The experimenter's **donning and doffing times** recorded for each participant and design are plotted in figure 8. In general, the donning of both designs took less than five minutes. The set-up time of both designs got faster with experience, as shown in the decreasing values in the left plot, from 135 to 74 seconds in the cuff and from 235 to 127 seconds in the bandage. On the right of figure 8, the donning and doffing times for the participants are also displayed. Donning and doffing the bandage design took, in

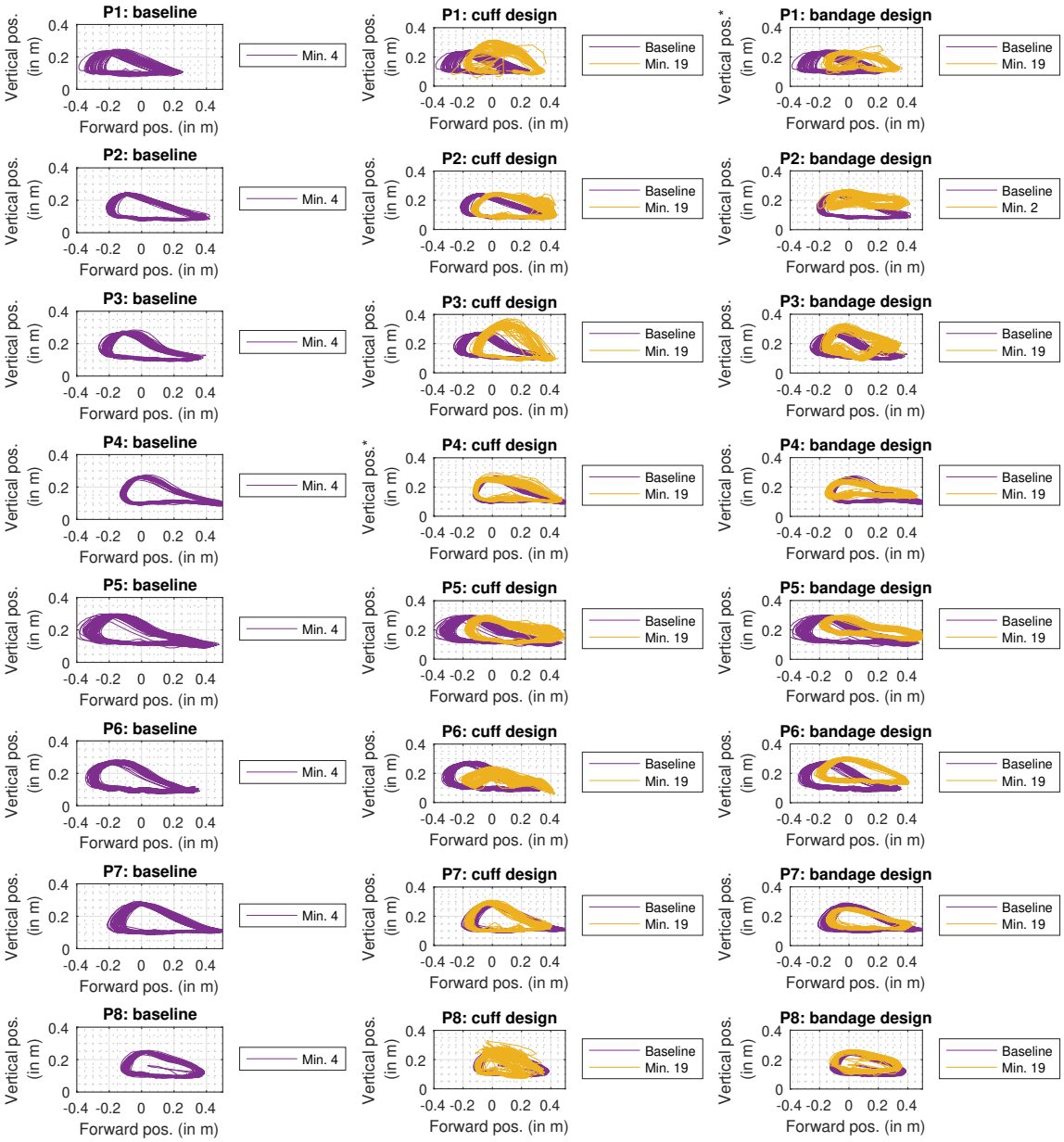


Fig. 5: Overview of the right ankle trajectories during baseline and during minute 19 with the bandage and cuff designs for each participant. The * (at P1 bandage and P4 cuff) indicates no data was available on the ankle height during mid-stance during that condition. Both times, the ankle height of the participant's baseline was used. Finally, the graph of P2 for the bandage design shows minute 2 instead of 19 because the bandage design got detached during walking in the third minute.

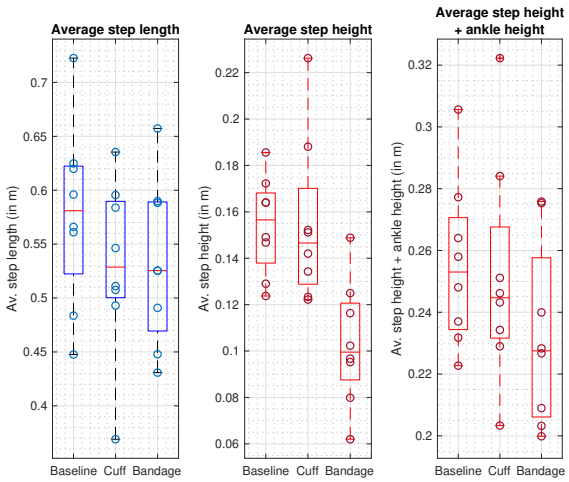


Fig. 6: Box plots of the average step length per participant, average step height, and the average step height + the ankle height for the different conditions. The median values of the box plots are shown in table II.

general, longer than the cuffs. On average, participants took 87 seconds to put on the cuff design and 217 seconds for the bandage design. For taking off the designs, the average time was 30 seconds for the cuff and 41 seconds for the bandage.

We observed **shear movement** with only the cuff design. The design slid up an average of 0.9 cm once the elastic cables of the MATE were connected to the design. This increased to 2.1 cm during the gait session. This data was obtained by measuring the height difference between the ground and the bottom of the right cuff design.

6) Notes from participants: Participant comments and quotes indicated that there were differing preferences in design. However, all participants noted that walking with the MATE became easier with experience. The bandage design provided a lifting effect while walking, and scuffing of the foot only occurred with the cuff design. With the cuff model, one's own walking pattern could often be established earlier, according to the participants. However, whether heel strike and toe-off could be accomplished better with the bandage or the cuff model depended on the individual. Generally, the bandage design was considered more controlling, particularly at the beginning of the gait session, while the cuff design was more guiding. The comments and quotes can be found in Appendix D-C2.

C. Design assessment

1) Feedback by Unit Infection Prevention (UNIP) and Create4Care: According to the interviewed UNIP and Create4Care employees, the cuff and bandage design used in the experiments could only be used once in a clinical setting due to their lack of washability. This limitation was attributed to several factors. Both designs had unsealed foam padding, which allows bacteria to infiltrate. Furthermore, the rough-textured lasting straps of the bandage design and the ridges and wrinkles in the synthetic leather covers of the cuff design were difficult to clean. Additionally, the aluminium base of the cuff did not tolerate cleaning products well, and the use of Velcro presented hygiene challenges. The employees also noted that the cuff design had an excessive number of parts, increasing the risk of losing components. More detailed information on the interviews with the UNIP and Create4Care employees can be found in Appendix E.

2) Observations during and after experiments: **Malfunctions with the bandage design.** The lower strap of the bandage design detached during the walking session for two participants: P2 after 2 minutes and P3 after 12.5 minutes of the 20-minute walking session. Both participants wore shoes with completely flat soles (P2: Adidas Superstar and P3: Nike Air Force), which caused the lower bandage straps to have no grip while walking. Additionally, P2 used a lower strap of the bandage model that was too large. In both cases, the participants continued walking without difficulty. However, P2 mentioned walking with a loose lower strap was less comfortable.

Damage on the designs. After the experiments, the non-Velcro part of the lower straps showed fraying around the back buckle of the bandage design. This mainly occurred because the Velcro regularly stuck to the lashing strap part of the straps. The cuff design did not suffer any noticeable damage during the experiments.

D. Case studies

1) Case study One: participant falls with the MATE and bandage design: Participant 8 (symbol in box plots: downward-pointing triangle) fell within the first minute while walking with the bandage model. A case study was conducted to gain a better understanding of what happened. In Appendix C-A, figure 22 displays the forces and kinematic data of the participant during the fall, along with the first minute of the baseline data and the data from the second trial of the walking session with the bandage model, where the front high cable was lengthened and the treadmill speed was reduced from 3.5 km/h to 2.5 km/h.

The case study sheds light on how the fall occurred. It was observed that just before the fall, the participant took longer steps, resulting in the legs being pulled forward more forcefully. This led to an excessively high forward force (235.6 N) on the right ankle, causing the leg to move forward and upward more vigorously than usual. As a result, when the left foot was about to push off the ground, the right leg was still unstable and positioned too far forward, preventing a successful push-off and causing a loss of control over the legs. The treadmill was immediately stopped after this happened, and the front high cable was elongated by 12 cm. Table VI displays the variance in cable pretension between the initial and subsequent trial of walking with the bandage design. Participant 8 was the sole individual to encounter this issue.

2) Case study Two: stroke rehabilitation physiotherapist evaluates MATE with both designs: The second case study was conducted with a stroke rehabilitation physiotherapist to evaluate the MATE and the designs. A condensed version of the experiment, i.e., 1 minute walking during baseline and 10 minutes with the MATE and each design, indicated similar

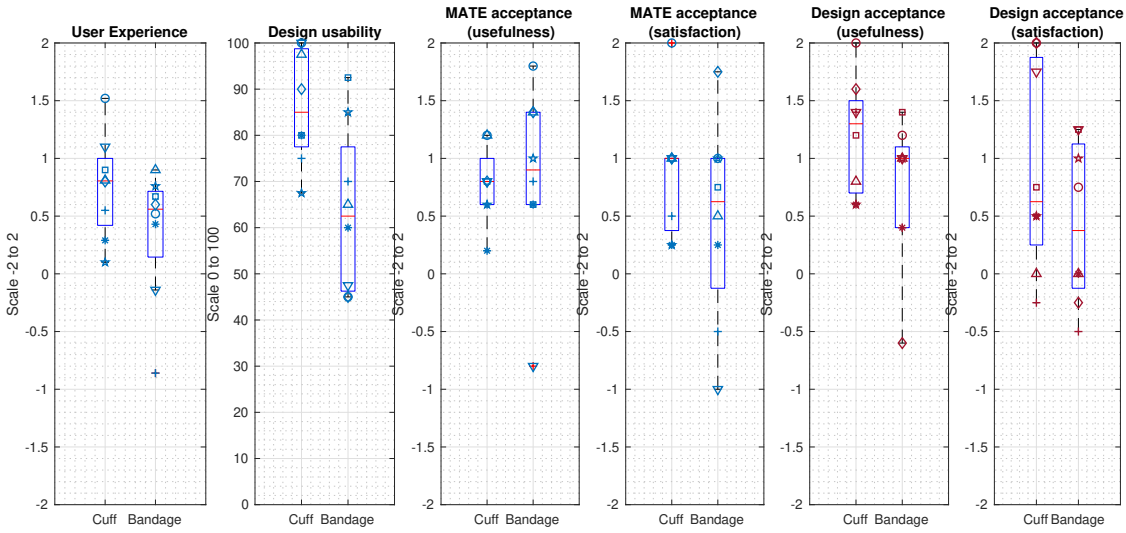


Fig. 7: The box plots show the questionnaire results of the User Experience, design Usability, MATE acceptance and design acceptance. Each participant is depicted with a personalised symbol. The values of the medians are reported in table III. Additionally, in Appendix D, more information can be found on the scores of the participants per questionnaire; tables XII and XIII, and figure 29 shows the enlarged versions of the box plots.

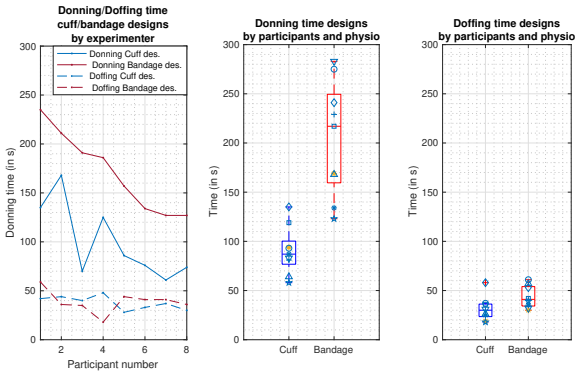


Fig. 8: The left plot shows an overview of the experimenter's donning and doffing times over time for each design. The middle and right plots, the box plots, show the donning and doffing times of the participants and the physiotherapist. Each participant's personalised symbol is used, and a yellow asterisk is added for the physiotherapist. The donning and doffing times are shown in tables XIII and XV, and an enlarged version of the plots in figures 30 and 31.

results to those observed in the participants. A second 10-minute trial with the bandage design and lengthened cables (at the physiotherapist's request) resulted in increased force on the design and reduced fluctuation amplitude compared to the earlier walking sessions with the shorter cables.

Importantly, the results show a clear preference for the bandage design from the questionnaires and the comments and observations. The physiotherapist mentioned that the bandage model better transfers forces from the MATE to the user and that only that design should be further developed. Finally, he stated, "With the cuff, walking was not energy-saving (you're trying to correct yourself more), and with the bandage, it was always energy-saving. But now that the cables have been adjusted, walking feels more natural. Own gait was interrupted last time (with the shorter cables), but now own gait can be completed much more.". The complete case study can be found in appendix C-B.

IV. Discussion

The experimental data is discussed according to the items on the list of requirements and the wish list. Furthermore, the study's limitations and future work are discussed.

A. *The attachment designs are an extension of the MATE*

Our first requirement was that the attachment design should be an extension of the MATE device, i.e., capable of comfortably transferring the pulling forces of the elastic cables to the human legs. It should also support the user's natural gait and increase the user's walking quantity and quality. This item is judged at the points listed below.

1) Supporting MATE's assistive purpose: The results reveal differences in how the legs were pulled based on attachment designs. The first difference arises from the type of connection between the human and the MATE. The cuff is connected around the legs just above the ankle. The bandage design is wrapped around the ankle, allowing for support to the natural swing movement of the foot. Participants' feedback indicates that the cuff design is perceived as more guiding, while the bandage model is considered as more controlling. The feeling of being controlled may be due to the higher forces pulling on the legs, as the data indicates that the bandage model exerted higher forces on the front compared to the cuff (on average). Furthermore, the amplitude of the force fluctuations during gait with the bandage design was higher than with the cuff, potentially contributing to the feeling of being controlled.

Differences in the pulling forces between designs could be attributed to the cables being connected to the legs at different positions and heights due to the differing attachment designs,

which may affect cable elongation and, consequently, pulling forces. These differences are reduced towards the end of the 20-minute walk with the bandage, in line with the participants' comments that, walking with the MATE becomes easier over time. It can be inferred that it takes time to adjust to the MATE, especially with the bandage design, as suggested by the changes in the kinematics between the second and 19th minutes of walking.

2) Range of motion and walking quality: The attachment designs must help support following the natural gait cycle as closely as possible without hindering the user's range of motion and adaptability. The results showed that walking with the cuff design is often perceived as easier and that the guiding forces allowed the user to stay closer to the natural gait cycle.

What is striking is that the kinematic data showed that the cuff design lifts the feet more. However, the participants indicated that they experienced a lifting effect from the bandage design and scuffing with the cuff (and that they may even trip because of this). Knowing that the ankle markers of the ZED mini shifted during the gait session and that a different ankle height was obtained in all conditions, it is likely that this kinematic data does not indicate the precise heights of the ankle. The kinematic data also showed that the step lengths when walking with the MATE are, in general, slightly shorter than when walking without. This also appeared to be the experience of participants. While the limited size of the treadmill was blamed by some participants, others noted that when making smaller steps, one could walk with the least resistance.

In addition, participants often mentioned the toe-off and heel strike to indicate how well they could follow their natural gait. A problem with both designs' toe-off was that the start of the swing phase, right after the toe-off, was cut short. During the heel strike in both designs, some participants experienced landing in the middle of the foot instead of in the heel. Also, participants mentioned it is harder to push/move the leg downwards for a heel strike when wearing the bandage design. This is important information, as stroke patients often have less muscle in their legs (than an unimpaired and healthy person) caused by muscle atrophy or hemi-paresis, and human tissues can be more flaccid for stroke patients, as mentioned in the *Discover* phase. Furthermore, depending on their natural gait pattern, participants noted that their legs are pulled to the sides (in both designs), creating a wider walking pattern, though one participant mentioned that his legs were pulled more to the inside. All these problems show that cable lengths and pulley positions should be adjusted per person's gait pattern so customised pulling forces provide the correct support.

Ultimately, whether participants can follow their natural gait depends on the circumstances. Some indicated that the cuff design allowed them to follow their natural gait pattern, and others questioned whether enough support from the MATE was being received. However, some participants commented that their gait pattern was controlled. Similar statements were made about the bandage design. Despite this, all participants agreed that walking becomes easier with experience. From their comments, it can be suggested that the longer they walk, the closer they get to their natural walking patterns.

Furthermore, participants' feedback revealed that the speed at which you walk can impact your experience with the MATE and how well you can execute your natural walking pattern. The closer you can match your natural gait, the better the support of the MATE, and the more enjoyable the experience. Participants mentioned that walking slowly is more challenging in both designs. However, some patients may need to start at a slower pace. The Lokomat evaluation form showed that patients' walking speeds ranged from 0 to 5 km/h, while participants in our experiment walked at speeds between 2.5 and 3.5 km/h.

Finally, during the first session with the MATE, wearing the cuff designs, the physiotherapist mentioned that he found walking with the MATE more pleasant than walking with the Lokomat, although not entirely comfortable. He noted that with the MATE, there was more freedom of movement and the ability to walk more naturally. He highlighted the advantage of being able to apply one's own gait and make compensations if necessary, unlike with the Lokomat, where movement is restricted. The physiotherapist emphasised the need for patients to practice using the MATE. Furthermore, he indicated a preference for the bandage model and mentioned that elongating the cables (when wearing the bandage model) made walking feel more natural. His natural gait was interrupted due to the shorter cables, but with the longer cables, he could complete his natural gait much better.

3) Design comfort: Based on the participants' comments on their gait pattern when walking with the MATE, user experience, and comfort, it appeared that staying close to one's natural walking pattern has the most impact on comfort. Furthermore, the tension in the cables could offer support and assist with the gait cycle. However, if the walking pattern is changed too much, participants might actively work against the pulling forces. Participants noted that walking became more comfortable when they were in the 'flow' or stopped resisting the forces.

As noted earlier, the cuff generally provided more guiding forces that made it easier to follow the natural gait cycle. This could explain why the user experience and 3 out of 4 acceptance questionnaire scores were higher with the cuff than with the bandage model. Conversely, the bandage model was perceived as more controlling, which might explain its higher score in the MATE acceptance (usefulness) compared to the cuff.

Furthermore, since participants related their ability to follow their natural gait cycle with comfort, it could mean that both designs are comfortable enough around the legs not to draw immediate attention. However, what did get a reaction was the fit of the designs. The cuff designs often slid up during walking, causing the fit to be too tight, and when the straps of the bandage model were not tight enough (or the shoe soles were completely flat), the lower strap would detach during use.

Additionally, just as there are variations in participants' walking patterns, there are also differences in comfort preferences. Some described the cuff as comfortable and feeling part of the body. However, others mentioned that the cuffs were more palpable and sometimes perceived as a warm band around the legs. When the cuff was worn directly on the

skin instead of over a sock, the participant noted that it felt sweaty but did not cause chafing. Some participants noted that walking began to feel heavier after about 10 minutes of wearing the cuff design, even though the design is lightweight. One participant suggested that this sensation could be due to the toe-off movement feeling different and proposed that his ankles enabled this change. Alternatively, it could be related to how the cuffs transfer forces to the body, which corresponds to the perpendicular forward movement of the leg, as stated by the physiotherapist in Case Study Two. The bandage model is also described as comfortable and more pleasant than the cuff. However, other participants mentioned the opposite or reported that they felt pressure on the ankles caused by the bandage straps. During one participant, the posterior buckle of the bandage design rested on the Achilles tendon instead of the shoe, causing slight irritation while walking.

B. Compatibility with prescribed AFO

This study conducted no tests using the attachment designs while wearing AFOs. Therefore, the only conclusions about whether the designs are suitable for all types of AFOs can be inferred from the physiotherapist's comments. The physiotherapist suggested that the cuff design might fit around an AFO, depending on the AFO type. He also suggested that patients might experience foot-dragging when wearing an AFO, which could pose a problem as foot scuffing occurred with the cuff design, as observed in healthy unimpaired individuals during the experiment. Regarding the bandage design, the physiotherapist indicated that some AFOs would work with the design while others would not. For example, the rear splint AFO might work, but the toe-off AFO probably does not.

C. User friendliness

1) Secure attachment: Both designs remained secure during the experiments. Even in the extreme scenarios of a detached lower strap of the bandage design and the fall of Participant 8 (Case Study One), the bandage design stayed in place around the ankle. Further research is necessary to determine what force would cause the designs to fail.

2) Accessibility and Usability: Even though both designs came in different sizes—small, medium, and large—to ensure a good fit for the wearer, fit issues occurred during the experiment. With the bandage design, the lower strap came off twice, and the cuff slid up. This should be further investigated and improved in future research, with special attention to the correct fit when wearing an AFO.

Regarding design usability, the cuff model performed better than the bandage model; however, this was not a unanimous finding. The experimenter's donning times revealed that both designs could be applied more quickly with practice. Participants specifically stated that they believed the bandage model became faster to use once they became familiar with how it worked. The physiotherapist commented that using different colours in the designs was a creative idea and emphasised the importance of making the product as user-friendly as possible. None of the participants provided feedback on whether the colours impacted their experience.

The cost per design can remain low since only inexpensive materials were utilised (straps, Velcro, synthetic leather, foam, buckles, and aluminium). The bandage design required fewer materials than the cuff design.

3) Short set-up time: On average, it took participants about 1.5 minutes to put on the cuff design and 3.5 minutes to put on the bandage design. It is worth noting that the time taken to put on the designs improved with practice. In the last trial, the experimenter took just over a minute to put on the cuff designs and just over 2 minutes to put on the bandage designs. When putting on the designs, including the body-weight support system and attaching the cables, neither design would have a set up in under 2 minutes for gait training. However, the cuff design's donning time showed the most promising result. Both designs were donned and doffed while sitting during the experiments, which were done without difficulty.

The protective pads of the bandage model need improvement to speed up donning and doffing. This would also prevent the separate pads from getting lost and enhance the design's appearance. The next step could involve incorporating the pads directly into the design of the bandage's straps. Furthermore, the closing flaps of the cuff's cover could be improved as the Velcro insides often stuck together or to the cuff, slowing down the process of putting the cuffs on and taking them off.

4) Quick-release option: There are different ways to do quick releases. The user can detach the four carabiners connecting the designs to the MATE and walk away with the attachment designs still attached. Furthermore, the designs are secured by Velcro straps that can be quickly and easily released. The physiotherapist could remove both designs within 30 seconds. Among the participants, the average time for removing the designs was 30 seconds for the cuff and 41 seconds for the bandage. These times can improve with experience, as shown by the experimenter's doffing times. Removing the cover for the cuff and the protection pads for the bandage design took the most time, and if this is improved, the designs might be taken off more quickly. Additionally, the designs' integrity was maintained during removal, so they might be donned even more quickly in an emergency.

When the body weight support system is still worn, the user cannot perform the quick-release options themselves. However, if it is already loose and the user is physically and mentally able to perform the actions, they can do it themselves.

D. Hygiene of the designs

From a hygiene perspective, both designs can only be used once. Additionally, both designs should undergo testing to assess their reaction to a 60-degree wash or disinfectant. Future research could also explore the use of a protective barrier between the human and the designs and the sanitation of both the designs and the human before use.

E. Suitability for patients in the early stages of rehabilitation

The use of the MATE early in the rehabilitation process should be reevaluated. Patients should be capable of walking independently to some extent before they can use the current MATE. The participants experienced difficulty walking in the

first stage of the gait sessions with both designs, so it is unlikely for early-stage walking patients to use the MATE in its current state. Additionally, the direction and level of pulling force patients need is still unknown. For instance, the physiotherapist remarked that when using the bandage model, his feet are lifted while walking with the MATE, but it would not be enough for someone with a drop foot.

F. Use without supervision

The cables regularly derailed from the pulleys during the experiments, requiring assistance to put them back in place. Additionally, supervision was necessary, especially at the beginning, to ensure the participants had control over their legs while walking with the MATE. If the participant demonstrated control, and there would be an automatic stop button for the treadmill in case of instability or a fall, then the MATE could be used in a situation requiring only mild supervision.

G. Blended care

1) Transportable: Both designs are lightweight and compact for easy transport between the patient's home and the rehabilitation centre.

2) User friendliness for patients and caretakers: During the experiments, the participants donned and doffed the designs themselves after just one demonstration. This could suggest that it is an easy task for healthy people, especially with some practice.

Whether patients can independently don or doff the designs is currently unexplored. No research was conducted on this, and the patient's level of disability would influence their ability to do so. However, according to the physiotherapist, patients would not put on the designs independently, as no physiotherapist would risk having a patient do this by themselves.

3) Instruction manual for the attachment designs: Neither of the designs comes with a manual. In future versions, a manual should be created with instructions on how to don and doff the design, connect it to the MATE, disassemble it for cleaning, and reassemble it. Adapted versions would likely be required for patients, depending on the level of disability. A size chart should also be included so users can easily find the right fit in the provided sizes.

H. Limitations

The study had limitations during the research phase, as many assumptions were based on interviews and visits to the selected rehabilitation centre. If a different rehabilitation centre had been chosen or a different physiotherapist had been interviewed, it could have led to different answers and assumptions, potentially affecting the attachment designs. Additionally, information about the MATE and/or the designs was not disclosed during the interviews, which meant that specific questions about gait rehabilitation and the purpose and operation of the MATE and the designs could not be asked, and no specific answers were provided.

Certain limitations regarding the experiment and its findings must be considered. It is important to note that the study solely involved healthy and unimpaired participants without an AFO, which may limit the direct applicability of the results to stroke patients and individuals who use an AFO when walking. Additionally, the study was conducted with a small number of participants. Furthermore, several uncontrolled factors were present in the experiment. For instance, individual footwear choices may have influenced performance, and the colour of participants' clothing could have affected the positioning of the ZED-mini markers. Moreover, the inconsistent placement of the ZED-mini camera may have influenced the placement of its markers and, consequently, the results. Additionally, the height of the ZED-mini ankle marker varied across different conditions and participants, potentially impacting the gait cycle results. It is also noteworthy that heel strike and toe-off actually refer to the ankle's position during these movements and that the average step length and height were determined using the right ankle marker. These limitations could be addressed with different sensing equipment, such as a traditional motion capture setup.

Furthermore, the lack of defined terms in the questionnaires could lead to subjective interpretations, impacting the consistency of responses among participants. The order of the designs during the experiment possibly influenced the results, as walking with the MATE appears to get easier with experience, even though we tried to minimise the level of influence by randomising the design order. Slight cable length changes during the gait sessions may have occurred, which could have influenced the force results. Finally, the exact centre of the treadmill was not explicitly indicated, but participants were guided by red tape on the sides. As a result, the middle location may vary between participants and during a gait session, likely influencing the measured pretension in the cables.

I. Future work

In the future, it is important to conduct research aimed at enhancing MATE's support to closely mimic natural gait while improving both the quality and quantity of walking and simultaneously increasing user comfort. One potential improvement could be to position the anterior attachment point closer to the toes, which may assist in preventing drop foot and provide better leg control. It is crucial to determine the optimal cable tension to provide support without excessively limiting leg movements or causing instability. Additionally, research should examine any variations in support and pulling force required to improve walking quality or quantity. A study could also investigate whether the attachment around both legs should be the same to maintain or improve symmetry or if there should be a different attachment around the patient's affected leg compared to the unaffected leg. Future studies should employ the C-mill to explore additional parameters for evaluating the participant's gait. Furthermore, since the MATE is attached to the ankle of the user, it is essential to investigate the behaviour of the ankle joint during walking with MATE and a specific attachment design, such as the ankle angle.

Additionally, it is necessary to increase the sample sizes and ensure a more diverse population to accurately determine

the impact of length, weight, muscle strength, gender, and age on the support needed from the MATE and the level of comfort it offers. In the future, conducting tests on stroke patients is important to focus our research on the specific support required for different disabilities. However, this will require initial testing to determine which AFOs can be used with the attachment design and to assess whether the design can be adapted to accommodate a wider range of AFOs. Subsequently, the next step could involve testing how the MATE and the C-mill function together for rehabilitation training.

In upcoming design iterations, it is crucial to focus on enhancing usability. This includes implementing built-in protection pads around the ankle and ensuring the bandage design is compatible with all types of shoes. Additionally, improvements in hygiene and user-friendliness are necessary, and tests must be conducted to determine the design's strength and the force at which it loosens or breaks.

In future research, direct engagement with participants to gather and analyse their feedback on gait adjustments and comfort is just as essential as analysing force and kinematic data. It is also imperative to involve stroke rehabilitation physiotherapists in optimising the MATE and the attachment design, especially as patient testing approaches.

V. Conclusion

Two different attachment designs for the MATE, a cuff and a bandage design, were developed and evaluated in this study. The design process followed the Double Diamond method, and the two designs served as connection points between the human and the MATE during experiments. The analysis revealed that the bandage design resulted in higher pulling forces on the legs and, according to the participants, made the support feel more controlling than the cuff design. The cuff design provides more guiding support, making it easier for participants to maintain their natural gait pattern. However, the new connection design should focus on enhancing the MATE's ability to support and improve the wearer's natural gait and enhancing walking quality and quantity by correcting the gait if necessary. Although the bandage design offers more feedback and support to the user than the cuff model (according to the participants), further iterations are required to improve gait support, comfort, and usability. In addition to enhancing the attachment design, adjustments to the pretension in the cables are necessary as they particularly affect the level of support and alteration of the gait and, therefore, highly influence comfort. After reviewing the results, ensuring that important requirements were met, and receiving the physiotherapist's recommendation to proceed with the bandage design, the preferred attachment design for the MATE was selected: the bandage model. This attachment design will facilitate a comfortable transfer of pulling forces from the elastic cables to the user without compromising the MATE's purpose, thus enabling the support of the user's natural gait and enhancing the quantity and quality of walking during gait rehabilitation for stroke patients.

References

[1] [Online]. Available: https://www.physio-pedia.com/Emerging_Technologies_in_Rehabilitation_for_Complex_Injuries_and_Conditions

[2] A. N. Malik, H. Tariq, A. Afridi, and F. A. Rathore, "Technological advancements in stroke rehabilitation," *JPMA. The Journal of the Pakistan Medical Association*, vol. 72, no. 8, p. 1672–1674, Aug 2022. [Online]. Available: https://pubmed.ncbi.nlm.nih.gov/36280946/

[3] [Online]. Available: https://www.hersenstichting.nl/hersenaandoeningen/beroerte/

[4] 2021. [Online]. Available: https://www.hartstichting.nl/hart-en-vaatziekten/cijfers-hart-en-vaatziekten

[5] G. Hope, "Subacute phase of stroke: Risks, complications, and treatment," Mar 2023. [Online]. Available: https://www.medicalnewstoday.com/articles/subacute-stroke#outlook

[6] Hocoma, "Lokomat® - hocoma," 2016. [Online]. Available: https://www.hocoma.com/solutions/lokomat/

[7] Motek, "C-mill." [Online]. Available: https://www.motekmedical.com/solution/c-mill/

[8] Aretech, "Zerog gait and balance system i aretech." [Online]. Available: https://www.aretechllc.com/products/zerog-gait-and-balance/

[9] MOTEK, "Rysen." [Online]. Available: https://www.motekmedical.com/solution/rysen/

[10] B. Haanen, "A mate for post-stroke gait rehabilitation; design optimization of a minimally actuated tendon-based gait rehabilitation device," Aug 2021.

[11] L. Kieft, "Design validation of the mate tendon-based passive device for post-stroke rehabilitation," Oct 2022.

[12] A.-M. Georgarakis, R. Stämpfli, P. Wolf, R. Riener, and J. E. Duarte, "A method for quantifying interaction forces in wearable robots," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*, 2018, pp. 789–794.

[13] B. Quinlivan, A. Asbeck, D. Wagner, T. Ranzani, S. Russo, and C. Walsh, "Force transfer characterization of a soft exosuit for gait assistance," 08 2015.

[14] Aguayo, "Double diamond methodology in user experience: Process and applications— aguayo's blog." [Online]. Available: https://aguayo.co/en/blog-aguayp-user-experience/double-diamond-methodology-user-experience/#:~:text=User%2DCentered%3A%20One%20of%20the

[15] J. O'Donoghue, "The double diamond framework for design thinking," Dec 2022. [Online]. Available: https://makeiterate.com/the-double-diamond-framework-for-design-thinking/#:~:text=It

[16] D. Council, "Framework for innovation." [Online]. Available: https://www.designcouncil.org.uk/our-resources/framework-for-innovation/#:~:text=The%20process%3A%20using%20the%20Double

[17] P. Rijndam, "Stroke rehabilitation on the lower body," Jun 2023.

[18] R. Revalidatie, *Lokomat Evaluatie Concept rapport*.

[19] T. I. D. Foundation, "What are personas?" [Online]. Available: https://www.interaction-design.org/literature/topics/personas#:~:text=Personas%20are%20fictional%20characters%2C%20which

[20] UEQ, "User experience questionnaire (ueq)," 2018. [Online]. Available: https://www.ueq-online.org/

[21] J. Van der Laan, A. Heino, and D. De Waard, "A simple procedure for the assessment of acceptance of advanced transport telematics." 1997.

[22] J. Brooke, "Sus: A quick and dirty usability scale," *Usability Eval. Ind.*, vol. 189, 11 1995.

[23] [Online]. Available: https://meetinstrumentenzorg.nl/instrumenten/functional-ambulation-classification/

[24] PhysioPedia, "Berg balance scale," 2009. [Online]. Available: https://www.physio-pedia.com/Berg_Balance_Scale

[25] B. Brody, "What meds do you need to take after a stroke?" Feb 2022. [Online]. Available: https://www.webmd.com/stroke/meds-after-stroke

[26] S. F. Kramer, S. H. Hung, and A. Brodtmann, "The impact of physical activity before and after stroke on stroke risk and recovery: a narrative review," *Current Neurology and Neuroscience Reports*, vol. 19, no. 6, Apr 2019.

[27] [Online]. Available: https://www.hersenletsel-uitleg.nl/gevolgen/cognitieve-gevolgen

[28] Y. J. Choo and M. C. Chang, "Effectiveness of an ankle–foot orthosis on walking in patients with stroke: a systematic review and meta-analysis," *Scientific Reports*, vol. 11, no. 1, Aug 2021.

[29] Y. J. Choo, J.-H. Kim, and M. C. Chang, "The prediction of need of using ankle-foot orthoses in stroke patients based on findings of a transcranial magnetic stimulation study," *Journal of integrative neuroscience*, vol. 20, no. 1, p. 119–119, Jan 2021.

[30] G. Paiz, "15 must-see user persona templates," Mar 2023. [Online]. Available: https://www.justinmind.com/blog/user-persona-templates/
[31] Y. Leng, G. Huang, L. Ma, Y. Qian, X. Chen, K. Zhang, and C. Fu, "A lightweight, integrated and portable force-controlled ankle exoskeleton for daily walking assistance," in *2021 27th International Conference on Mechatronics and Machine Vision in Practice (M2VIP)*, 2021, pp. 42–47.
[32] A.-M. Georgarakis, R. Stämpfli, P. Wolf, R. Riener, and J. E. Duarte, "A method for quantifying interaction forces in wearable robots," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*. IEEE, 2018, pp. 789–794.
[33] R. Alicea, M. Xiloyannis, D. Chiaradia, M. Barsotti, A. Frisoli, and L. Masia, "A soft, synergy-based robotic glove for grasping assistance," *Wearable Technologies*, vol. 2, 2021.
[34] O. device maker, "Interview with orthopaedic device maker about afos," Jul 2023.
[35] [Online]. Available: https://orthoticsplus.com.au/orthotics/ankle-foot-orthoses-afo/
[36] [Online]. Available: https://www.klimwinkel.nl/petzl-irvis/flexlock-2023
[37] N. Ize, "Camjam xt spanhaak zwart l — nite ize nederland." [Online]. Available: https://www.niteize.nl/p/camjam-xt-spanhaak-zwart-l-ni-ncjla-01-r3
[38] [Online]. Available: https://www.erasmusmc.nl/nl-nl/patientenzorg/specialismen/infectiepreventie#04fdf454-0dce-482d-a76c-10b60eff90d1
[39] [Online]. Available: https://www.erasmusmc.nl/nl-nl/patientenzorg/create4care#9773ed8a-7df2-4920-a949-19ed7654a57a
[40] [Online]. Available: https://store.stereolabs.com/products/zed-mini
[41] StereoLabs, "Body tracking overview - stereolabs." [Online]. Available: https://www.stereolabs.com/docs/body-tracking
[42] Stereolabs_cameraCalibration, "Camera calibration - stereolabs." [Online]. Available: https://www.stereolabs.com/docs/video/camera-calibration
[43] K. Betteridge, "What every client should know about sus scores — bentley university." [Online]. Available: https://www.bentley.edu/centers/user-experience-center/what-every-client-should-know-about-sus-scores#:~:text=The%20SUS%20score%20is%20the
[44] J. Sauro, "Measuring usability with the system usability scale (sus)," May 2016. [Online]. Available: https://www.userfocus.co.uk/articles/measuring-usability-with-the-SUS.html#:~:text=A%20SUS%20score%20above%20a
[45] RELAB, "Usability evaluation database," 2024. [Online]. Available: https://www.usabilitytoolbox.ch/database/#results

Supplementary Material

Design and evaluation of the connecting element between a human and the MATE, a tendon-based rehabilitation device for the lower body

APPENDIX A
DOUBLE DIAMOND METHOD: EXTRA MATERIAL

A. Discover

1) Core personas: During the *Discover* phase in the Double Diamond framework, core personas are created that represent the community one is designing for. Their struggles, wishes, and needs for a future design are presented and prioritised. For this study, two personas were created, Paul and Susie; see figure 9. This was done via online research on stroke patients, an interview with a physiotherapist of a rehabilitation centre specialised in stroke rehabilitation, and observations of stroke patients during rehabilitation physio sessions. Although creating personas is a valuable way to get to know the user, it is important to state that every patient is different, has a different set of disabilities, and has a different recovery after their stroke, which means that comparing them cannot really be done. Susie and Paul each have a range of common and impactful disabilities, issues, and struggles that people experience during stroke rehabilitation to cover as many important aspects as possible. Below, the information that inspired the core personas is given.

Gender: When creating the personas, a man and a woman were selected. During the interview [17], it emerged that rehabilitation devices can be more or less comfortable depending on gender. For women, when waistbands are used, the breasts can get in the way when this waistband shifts upwards, which can create a huge and uncomfortable push-up effect. Men, more often than women, report that leg straps in the groin cause a lot of discomfort. Both the waistband and leg straps are present in a body-weight support system/safety harness.

Age: The personas are also in a different age category. This was done because it was mentioned during the interview that young people are generally a bit more adaptable and quicker to adapt than older people. In addition, it could be that older people were already taking a higher dose of blood thinners, which could cause them to bruise faster.

Date of Stroke: In both the personas, the patient is in the sub-acute phase after the stroke. This phase, which starts a week after the stroke till six months post-stroke, is called the "prime time for rehabilitation". During this time, the brain and body are adjusting and reconstructing to the impact of the stroke [5].

The stroke date does not indicate whether a patient should be walking at a certain level. The physiotherapist decides if and when the patient is ready to start walking, looking at their ability to stand, foot clearance, absence of tilt in the foot when putting on the ground, and symmetry. With the FAC-score (Functional Ambulation Category) the level of independence in walking is assessed [23], and with the Berg Balance Scale (BBS) the the patient's balance ability during tasks is evaluated [24].

Medication: Susie and Paul both take blood thinners to prevent future cluttering of the blood [25]. This is important to know because, as mentioned above, due to the blood thinners brushing the patient's body can happen more easily.

Living situation: Showing two different living situations is important because once the patient starts to go home more often, more responsibility is given to the family and/or friends of the patient. This also means that when certain tasks and exercises need to be done at home, family members should know how to help the patient and possibly attach certain assistive devices the correct way.

Openness to technology: Embracing technology in healthcare is important as rehabilitation and technology increasingly intertwine [1]. Not only are robotic rehabilitation devices becoming more popular, but technology also brings advantages at home. 'Smart home' devices could take away some of the physical tasks when a patient returns home. Additionally, through technology, blended care is becoming a more acceptable option in stroke rehabilitation.

Fitness and lifestyle before stroke: It is important to know the fitness of the patient before and after their stroke, as physical activity (PA) can reduce the risk of a (recurrent) stroke [26]. The same study states that "the intensity of pre- and post-stroke PA and exercise is a key factor with higher intensities leading to greater benefits, including improvement in fitness. Having low fitness levels is an independent predictor for increased risk of stroke" [26].

Type of disability: As mentioned, every patient and their rehabilitation process are different. However, one can distinguish between cognitive disability and physical disability. During rehabilitation with a patient suffering cognitive impairment, the

level of stimuli and the level of aphasia and fractality of the exercise should be taken into account. This should also be considered when giving instructions. Another form is 'neglect', suffering impaired perception of the affected side [17]. When focusing on gait cycle rehabilitation, the degree of hemi-paralyses or (hemi-)paresis, so the degree of muscle failure has the most influence. However, many other after-stroke effects influence gait training [17], like drop foot, a stiff knee, muscle spasticity, the level of stimuli a patient can handle (e.g. having trouble with talking and walking at the same time), or an impaired perception of the affected side of the body.

Notes for rehabilitation: Besides the disabilities mentioned in the '*type of disability*' paragraph, other issues can occur during physio sessions that influence the performance. Some of the issues that cause problems or difficulties during gait training are drop foot, spasticity, and stiff knee (all influencing the natural walking movement). During gait training, starting the initial swing costs the most energy, and in the stance phase, knee stability is a point of attention. Some patients develop the so-called 'stiff knee gait' caused by a lack of hip and knee flexion, influencing the start of the swing phase. And, the over-extension of the knee in the stance phase creates instability problems. In the long term, constantly overextending the knee can eventually start causing knee problems. In general, for lower extremity rehabilitation, it is mainly the hip region that tends to get painful since patients have to use a lot of force to move the leg forward [17]. And, sometimes being tired is enough to influence the level of performance during an exercise. Foot dragging, often, is the first signal that a patient is getting tired, lifting the leg then starts getting harder[17].

Another important factor that influences performance is 'attention to the task.' And, many different reasons could influence one's attention span. With the help of the cognitive pyramid, insight can be given into which cognitive function disorders belong at each level of the pyramid, and how the different levels influence each other [27][17].

The last point of attention that could cause issues in physio sessions is rehab materials, such as the waistband and straps in the groin area already mentioned above. In addition, when using straps around the legs, for body-weight support, possible muscle atrophy of the patient must be taken into account. Especially when the straps make use of the tapered shape of the leg. One should prevent, when a patient with a lot of muscle atrophy (where the hip is as thick as the knee) wears a strap placed above the knee, it starts sliding upwards during use instead of sticking to the thinner part of the leg [17].

Use of an AFO: An ankle-foot orthosis (AFO) can enhance a person's gait stability after a stroke, even as the walking speed, step length and cadence [28]. It is a popular tool during stroke rehabilitation and at discharge still 39% of stroke patients need an AFO [29]. The AFO does not need to be worn all the time during rehabilitation. It depends on how quickly and well a patient's rehabilitation is going. For example, some only wear their AFO when they go for a walk outside, walk on uneven surfaces, or when they go on longer walks. Walking without their AFO indoors allows them to practice in a low-key way.

Oedema formation in legs and pressure sores: When designing an attachment to the legs, it is important to take into account ailments that can affect the design's impact on the patient, like muscle atrophy mentioned above. Stroke patients with a paralytic leg or a paretic leg often experience oedema formation in the legs, which can cause a size difference between the affected and unaffected leg [17].

The location of pressure sores on the legs depends on i.a. the type of AFO patients wear and the degree of oedema in the legs. Possible places could be at the rim of the footplate of the AFO, or the heel depending on the design of the AFO. Because AFOs are customised pressure sores do not often occur on the calves of the patient. However, what can happen are light shear forces where the edges of the AFO touch the calves. People with a shoe provision often have pressure points around the ankles, unlike people who only wear an AFO. Finally, when a patient suffers from Hallux Valgus, pressure sores could form on the cusp of the big toe [17].

What motivates the patients: What motivates is when patients can see that they have made progress. When they are allowed to exercise harder or have achieved a new milestone, such as being able to start walking training, their first steps, or when they can go to the toilet by themselves. Furthermore, it is important to make exercises challenging, but easy enough that patients can still 'score' and get lots of positive feedback [17].

Challenges: Patients get easily distracted, see paragraph '*notes for rehabilitation*' for information about the cognitive pyramid. Another challenge is 'frustration,' which could be caused by only occasionally succeeding at a task due to fatigue (limited workload). Or when a patient is not succeeding in a task, because they do not have the functionality to do it. Not being able to influence this can be frustrating for patients [17].

Fears: In the interview, it was said that when it comes to training with the gait rehabilitation devices, patients can be afraid of falling. Also, the chance of damaging the body is mentioned as a fear. Although, people are not afraid that the attachments of the rehabilitation devices will hurt them. Lastly, the moment when the patient has to perform the task alone, without a physiotherapist present, can be experienced as tensive.



(a) Persona One

(b) Persona Two

Fig. 9: Susie and Paul are the two core personas created for this study. They represent the users one is designing for in this study. Their struggles, needs, and wishes have a great influence on the design process. (Persona head shots [30].)

2) Physiotherapist journey map: As mentioned, understanding the user's wishes and needs is a key aspect of the Double Diamond design process. Earlier the core personas are specified, in which the patient is represented. However, the struggles, needs, and wishes of the treating physiotherapist are not to be ignored in this process. In this sub-chapter, an overview is created of the journey of a stroke rehabilitation physiotherapist during a session. Figure 10 is constructed via an interview with a physiotherapist from a stroke rehabilitation centre [17]. The dark blue circles show the extra steps a physiotherapist must take when using a robotic device during a session. The light blue steps are to be completed with and without a robotic device. Below an elaboration is given on the struggles during this journey.

Struggles and wishes of the physiotherapist, during a physio session [17]:

- The most crucial struggle and wish is to have a quick setup time. Boot up rapidly, connect swiftly, and work efficiently.
- Another wish and current struggle is that patients should be able to use the robotic device without supervision, allowing one physiotherapist to treat multiple patients simultaneously in the same room. Requiring constant supervision for safety would be a significant limitation.
- The interview reveals that most frustrations are related to software issues. Many productivity problems persist, such as connection and start-up issues, and the user-friendliness of the technology also needs improvement. While some problems are solely caused by the system, it is also common for these issues to be a combination of system flaws and the therapists' digital skills.
- Also, most systems currently cannot be put on properly in a sitting position. They can be put on, but they don't fit properly. As a result, when a patient is unable to stand, they are put on the treatment bed when putting the system on. A wish is to be able to properly put on a system while the patient is sitting.
- Finally, more emphasis should be placed on blended care, improving the collaboration between technology and care (also see the second bullet point), and online rehabilitation for patients to do exercises from home.

B. Define

1) List of requirements and Wish list: In the sub-chapter *Define*, all problems found in the *Discover* phase are prioritised and the biggest problem and the main objective are identified. A list of requirements and a wish list are compiled to distinguish between various features for the future design. The requirements list includes all the features necessary to achieve the main objective, while the wish list includes features desired but not crucial for achieving this main goal. In figure 11 the list of requirements and the wish list can be found, and below elaborations on all listed design features are stated.

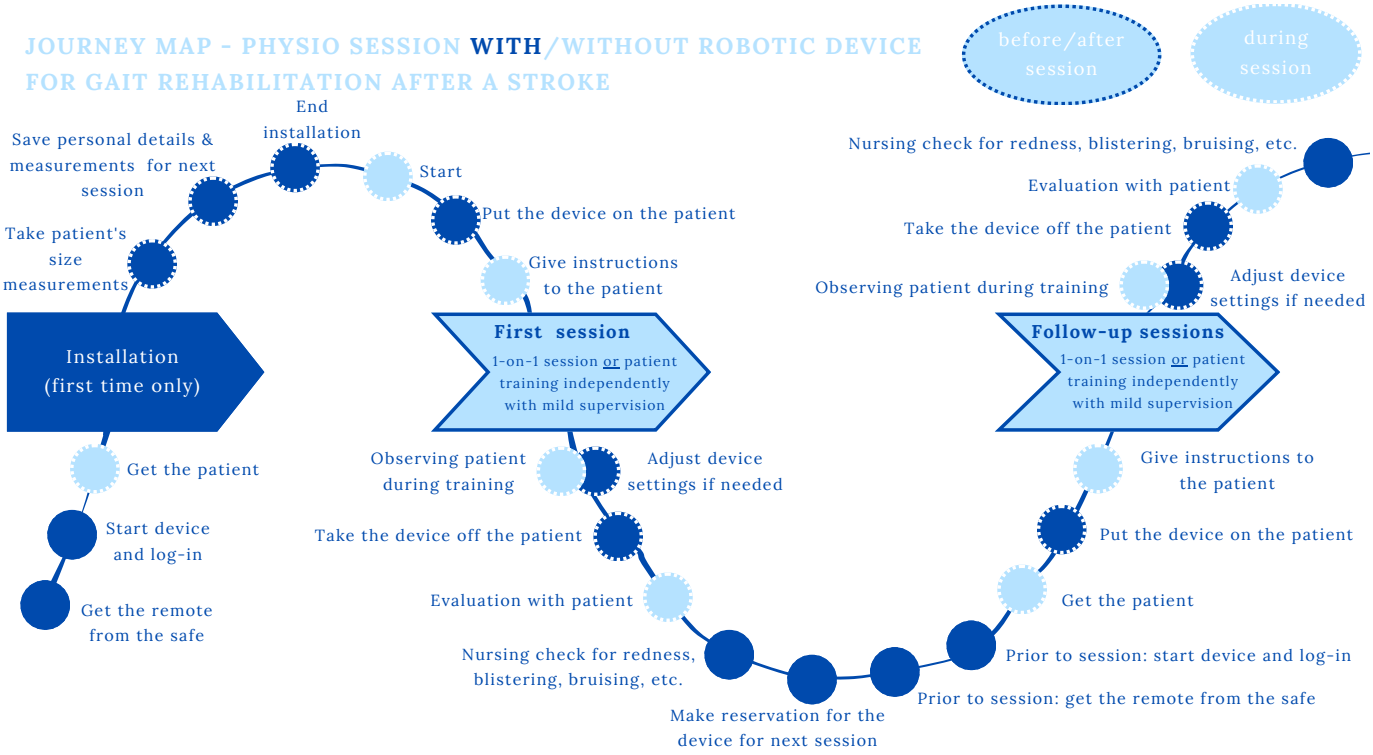


Fig. 10: Journey map of the physiotherapist at a stroke rehabilitation centre. This figure represents the journey of a physiotherapist during a session with a stroke patient when using a robotic rehabilitation device and when this is not used. It is important to notice the number of extra steps it takes to work with robotic rehabilitation devices during a 30-minute physio session.

The design must help increase the purpose of the MATE: The primary purpose of this connecting design is to attach the MATE to the human body. It is important to secure a comfortable connection while still achieving the goal of the MATE. Proper use of the device should lead to an improvement in both the quantity and quality of walking. However, as mentioned in the *Discover* phase, the initial focus should be on improving the quantity of walking. Once the patient experiences functional recovery, the quality of walking becomes more important.

The design must be comfortable: The interview revealed that patients consider comfort to be very important. Sometimes, they want to stop an exercise because the device or attachment does not fit comfortably. Therefore, it is essential that the to-be-created attachment design does not cause any discomfort. This feature was chosen as the most important from the patients' point of view.

A physiotherapy session is typically 30 minutes per patient, but the duration may vary based on the patient's stamina. Excluding the time for putting on and taking off the equipment, a walking training session lasts about 25 minutes [17].

The design must not affect the patient's range of motion: In the *Discover* phase, it was noted that one of the limitations of the Lokomat is its restricted range of motion. Therefore, the future interface between the MATE and the human should enable a full range of motion to allow patients to compensate for their gait motion during walking and provide room for patients to make mistakes and allow for a natural trial-and-error process.

The interview revealed that the most crucial aspect of walking support is to not hinder the patient's adaptability. Therefore, this feature was chosen as the top priority by the physiotherapist from the list of requirements.

The design must not affect the walking quality negatively: The attachment design should not, in any way, have a negative effect on the gait pattern of the patient or their (gait) rehabilitation process.

Optimising quality means ensuring that the functional recovery occurring at the muscle level can be utilised as much and as effectively as possible. This allows the patient to walk with the highest quality possible, as walking with better quality consumes less energy. In other words, the aim is to follow the normal gait pattern as optimally as possible [17].

The design must allow patients to wear their own AFO, shoe facility, or brace without limiting its function: An AFO is a customised ankle-foot orthosis, used to improve gait and gait stability, see the *Use of an AFO* paragraph. The use of an AFO is often a part of the gait rehabilitation process. Patients initially walk with an AFO, but as they regain function, they begin

practising walking without it. There may be a period where patients wear the AFO in daily life but train without it during physiotherapy. So it's common that sometimes people wear an AFO and sometimes they don't. Therefore, the design must 1) allow patients to wear their own AFO without limiting the function of the AFO, and 2) perform the same when patients do not wear an AFO. It is important to note that, in this stage of the design process, neither the MATE nor the attachment designs are able to replace the role of the AFO.

The design must be attached securely: The design must not break during use, and also provide the patient with a sense of security.

The design must be accessible: As mentioned earlier, one cannot compare one stroke patient with the other, because every patient is different and has a set of different disabilities. Therefore, it is important to provide a wide range of sizes to make sure each patient finds one that fits perfectly around their legs and AFO. Additionally, the design needs to be affordable, so the MATE is accessible to everyone who needs it.

The design must have a 2-minute set-up time (max.): The number one struggle of a physiotherapist is not having a quick set-up time, see '*struggles and wishes of the physiotherapist, during a physio session*'. In the interview, it was concluded that having a set-up time of 2 minutes is acceptable, from zero to gait training [17].

The design must allow to be attached while the patient is sitting down: This was one of the wishes in the paragraph about '*struggles and wishes of the physiotherapist, during a physio session*.'

User-friendly: Having an easy user interface that allows for quick and accurate application of the device will reduce setup time and eliminate the need for additional device training.

The design must have an option for a quick release: In case of an emergency, there must be an option to release the patient from the attachment designs within 30 seconds [17].

The design must be disinfectable: Hygiene is crucial, especially in healthcare. Therefore, the design must be easily disinfected by either washing it in a washing machine at 60 degrees Celsius or by creating a design that can be wiped with a disinfectant cloth.

The design may be usable in the early stages of stroke rehabilitation: This statement is included in the wish list because it depends on two things. Firstly, the physiotherapist decides if the patient is ready to start doing walking exercises, looking at test results and rehab progress, as mentioned in the 'Personas' sub-chapter. Secondly, before any conclusions can be drawn about whether the design provides the correct support and the right amount of support, tests need to be conducted and analyses need to be carried out. It is highly likely that more iterations and tests are necessary before this goal can be achieved.

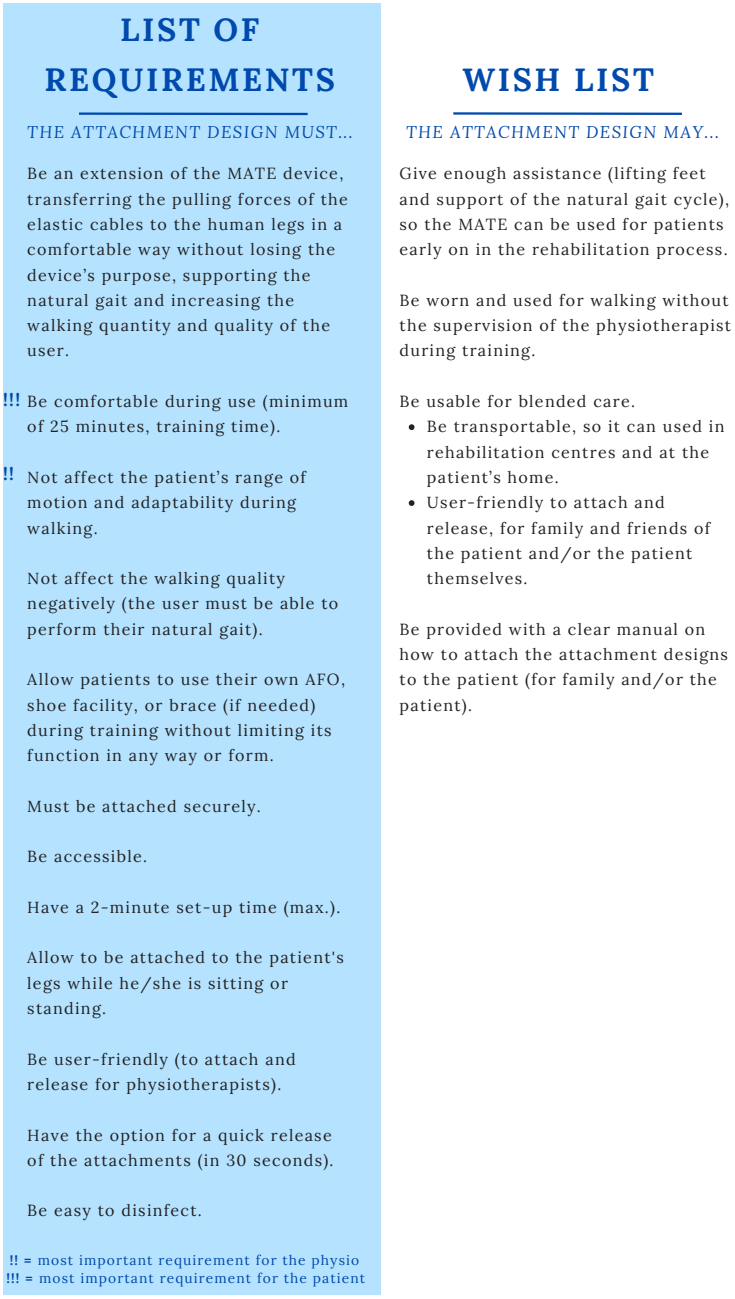


Fig. 11: List of requirements and wish list for the future MATE attachment design, based on the findings in the '*Discover*' and '*Define*' phase.

The design may be used without supervision: Using the design without the supervision of the physiotherapist is put on the wish list for now. This is the first attachment design that allows for walking with the MATE for a longer time in a comfortable way. As listed in 'requirements', the goal is that the attachment designs are an extension of the MATE and comfortably transfer the pulling forces while still supporting the natural gait. However, because the design of the MATE, the design of the attachments, and the combination of those two are in the early stages of testing and development, supervision of the user is still essential. In the future, this feature must be put in the list of requirements instead of the wish list.

The design may be used for blended care: If blended care, where the patient can also do more exercises from home, is going to be a part of the near future, the following options will need to be taken into account. The design must be easily transportable from the patient's home to the physiotherapist and back again (if the design is customised). Additionally, putting on the design must be an easy task so the patient's caretakers or the patient themselves can do it.

The design may be provided with a clear manual: If the design is intended for use in blended care, as mentioned in the paragraph above, it should include a very clear manual. Priority should be given to informing the patient's caregivers effectively, as well as informing and explaining the manual to the patient. When creating a manual for the caregivers, it's important to consider their struggles, insecurities, and questions. When designing for the patient, it's important to consult with a physiotherapist and research the effects of cognitive load and the cognitive pyramid, as explained earlier. Also, one should consider the effects of the patient's disabilities on securing the attachment design, e.g. possible hemi-paresis, level of neglect of one side of the body, or the level of aphasia of fractality.

C. Develop

In the *Development* phase, different solutions and ideas are explored to solve the problem defined in the previous chapter. In this chapter of the appendix, more elaboration is given on all the explored fields, steps, and choices to eventually create the final design concepts for the next phase, *Deliver*.

1) Design choices of the attachment design that will influence the features on the list of requirements: Before starting the design process of the attachment design, some clarification needs to be made on the role of the attachments in achieving the features on the list of requirements.

Supporting the natural gait and increasing walking quality and quantity: Besides supporting the swing phase initiation, as mentioned in the subchapter *Define*, the MATE design must support other steps in the gait cycle as well. Below are some points the attachments should include to get to the goal of improving the natural gait.

- *Support with initiating the swing phase.* The design should include an option to lift the foot at the toe-off.
- *Supporting completing heel strike.* The design should include an option to keep the ankle dorsiflexed during the swing phase. Wearing an AFO during gait training could help with this. Also, help guide the foot towards the correct place to heel strike.
- *Support stabilising mid-stance.* The design must not block ankle movement in any way. Also, an AFO could help with gait stability.
- *Supporting bending the knee during the swing phase.* Again, the design must not block this movement. The MATE and design might help bend the knee by lifting the foot at the toe-off.

Additionally, the position of the attachment on the legs and the position of the cables on the attachment are design choices that will influence the support of the natural gait as well.

- *Position of the attachment on the legs.* During the interview with the physiotherapist, he was asked where he would place the attachments if he wanted to control someone's leg during gait training. If a patient walks with their AFO, the placement would be around the ankle and, without AFO, around the anterior arch of the foot, [17]. Also, in literature, the ankle joint is successfully chosen as an effective location for applying for assistance [31].
- *Position of the cables on the attachment.* In the previous attachment design, the elastic cables were positioned on the sides of the leg, leading to shear stress between the wearer and the design during use. After conducting force calculations, see Appendix B; it became evident that the forces in the front-back (anterior-posterior) direction are much higher than those in the side-to-side (sagittal) and up-and-down (transverse) directions. Therefore, it is more effective to group the cables at the front and connect them to a point at the front of the design while also connecting the grouped rear cables to a point at the back of the design. Only having two points connecting the elastic cables to the attachments in the same line as the direction of the highest force can reduce the rotation of the design during use and, therefore, minimise shear stress.

Allow full range of motion and adaptability: The MATE must allow the patient to be in control of their movements while giving support and guidance towards an improved gait. The attachment designs must comfortably transfer the pulling forces

to the user.

Ensuring comfort of the patient: Comfort is a wide concept influenced by many factors. When one wants to influence comfort, multiple design features should be considered. The factors that influence comfort in robotic wearables are:

- *weight.* One wants the design to be light so it does not impose extra effort on the wearer or negatively influence gait motion.
- *Size.* The design should be small, minimising interference with the mobility and the gait cycle.
- *Synergistic interaction with the MATE.* Comfort, safety, and performance of the MATE benefit from synergistic interaction between the wearer and the device [13][32][33]. Besides guiding the user into the correct gait cycle, the MATE's level of assistance must match the walking pattern and kinematics of the wearer. So, higher pulling forces do not necessarily ensure better support or results. The design will have to transmit the forces gradually, naturally, and securely to ensure that the attachment is not the limiting factor. For the design process, this means providing design features that will transmit every pulling force in every direction in the way the support is intended.
- *Pressure distribution on the place of the attachment.* Uniform pressure distribution is desired as it minimises the chance of high-pressure points and restricted blood flow.
- *Shear stress between design and wearer.* Shear stress can be reduced by selecting the correct force path in the design.
- *Fit of the design.* A good fit lowers the chance of shear stress between the design and the wearer or restricted blood flow. Also, it helps optimise the synergistic interaction of the device and the human. So, the attachment design should be re-adjustable or come in different sizes.
- *Material specifications of the design, e.g. padding and straps.* The design materials should not irritate the user's skin and ensure that relative motion between the user and the design does not cause a problem.

Design can be worn with and without AFO & the AFO does not lose function or comfort when the attachment design is connected to the leg: To gain a better understanding of the functioning of an AFO and what external influences affect its functioning, an interview was conducted with an orthopaedic device maker at a stroke rehabilitation centre. Factors and possible design features that influence the functioning of an AFO are:

(Side note: the MATE was not explained during the interview, and the design concepts of the attachments were not presented.)

- *Heel height influences the functioning of the AFO.* The heel height of the shoe affects the functioning of the AFO, as the customised AFO then pushes the leg at a different angle. With a raised heel, the AFO will push the calf more forward than normal, causing more pressure on the calf, which can cause knee instability [34].
- *Shoe height affects AFO function.* A high boot can restrict AFO function because it makes it much harder to move the ankle. Whether a patient walks better in mountain boots or sports shoes is person-specific. So, the type of shoe of the patient influences the functioning of the AFO.
  The weakest point of a back splint AFO is at the Achilles tendon, that's where most of the force is concentrated. However, suppose a high mountaineering shoe is worn. In that case, it becomes less easy to bend at that part because the shoe will counteract it, so less force is exerted on that part of the AFO because the shoe is more rigid and, therefore, stiffens the stand phase during the gait cycle [34].
- *Attachments to the leg while wearing an AFO.* If a cuff or strap is placed around the ankle or lower leg while wearing an AFO (so that the strap goes around the leg and the AFO), and a pulling force is applied, these extra forces around the lower leg will not considerably affect the AFO, as the AFO already fits securely against the leg during use. A strap around the foot's instep helps the AFO stay in the shoe better, making it a favourable place for an attachment. As long as the cables of the MATE are not connected to the AFO but to a separate part that wraps around the shoe or leg (the attachment design), pulling forces of the cables on the connection points on the attachment designs, both at the front and back, are unlikely to have negative effects on the functioning of the AFO. Additionally, AFOs are sturdy enough to be used for sports [34].
- *Preferred location of a connection to the leg which has the least impact on the AFO.* If the design is put around the foot, it also has to go into the shoe, so placing the attachment on the leg is more comfortable for the patient (and gives less chance of pressure sores on the foot). However, if the connection goes around the shoe, it is more hygienic than around the leg, as the patient's skin will not be touched [34].
- *Issues to consider when designing something that connects to the patient's feet or lower legs, when wearing an AFO.* The connection should not exert too much pressure on the device, because if it exerts a lot of pressure on the AFO, it also exerts pressure on the body and increases the chance of getting pressure sores. It is better to follow the body than to press something against it. Also, the attachment must not be too tight around the body to maintain free blood circulation. During movement, the AFO shifts a bit against the leg regardless, some people solve this by wearing high socks under the AFO. The AFO generally reaches up to two fingers below the fibula head of the knee. Furthermore, depending on

the size and weight of a person, an AFO can change the thickness of the patient's leg, as larger people need thicker material for the AFO. However, the average thickness of an AFO is 4 mm [34].

As mentioned, the patients should be able to wear their own AFO, shoe facility, or brace while walking with the MATE. However, there is a wide array of AFOs, as different kinds of AFOs give different support, e.g. a leather inner shoe secures the foot and ankle much more than a back splint AFO. Usually, stroke patients only have a back splint AFO [34]. Different types of AFOs are the Ground Reaction AFO, Leafspring AFO, Solid AFO, Jointed AFO, and Posterior Dynamic Element AFO [35]. In addition to the differences in function and support, these AFOS also have different designs. For example, an AFO can be placed at the back of the leg, from the heel up the calves, but another can rotate around the leg. These differences must all be taken into account when creating the new attachment designs.

User-friendly interface: Design features that will help achieve a user-friendly interface start with design simplicity. Creating a simple design with a small amount of different parts will make it easier to use and easy to understand, and these characteristics lead to a quick and easy setup.

Ensuring hygiene of the design: When ensuring the hygiene of a design, several choices can be made. Firstly, it is important to create a design without small notches or a lot of seams. Additionally, materials should be chosen that can withstand a 60-degree Celsius wash or are made of disinfectant-compatible material. Moreover, durable materials should be prioritised to ensure high re-usability and avoid the need for a new design after only a few uses. Furthermore, the decision to customise each attachment design or create a general and interchangeable design between patients (which is more sustainable) affects the degree and intensity of disinfection required after use. Finally, the design decision regarding whether the design is worn around the shoes, on clothing, or on the patient's skin also influences the degree of disinfection required after use.

Ensuring safety during use: The design strength is determined by both the shape and material of the design. Therefore, the design process must consider the optimal path of the forces exerted on the design. The design should be strong but also have a degree of flexibility to prevent breakage. Additionally, materials should be selected for their strength and resistance to rapid wear or loss of function.

2) Mind map: A mind map is created in this design process to explore similar problems in other fields and industries (nature, sports, fashion, healthcare, and engineering), see figure 12. Special focus is on how the devices are connected to the human or how two objects are connected. Also, the focus lies on the material of these connections. Finally, matching solutions in different industries are given a closer look at the reason they were used in that particular problem, looking for safety reasons, user-friendly designs, and interest in elevating performance. For example, a harness design is often used when a firm connection and safety are needed.

Across all different fields in the mind map, the most common types of connections are:

- Wrapping one object around another, e.g. using a harness, a cuff, or a cover.
- Hooking one object into another, in case of more permanent or stronger connections.
- Pass one object through another, in case gaps are present in an object.

And, within these different types of connections the most used connecting/fastening systems are:

- Straps: elastic, Velcro, and rubber (e.g. seat-belt)
- A knot joint of e.g. ropes, shoe laces, or tangling of material.
- Buckles.
- High friction between material(s).
- Locking systems e.g. clips, click bindings or a twist lock.

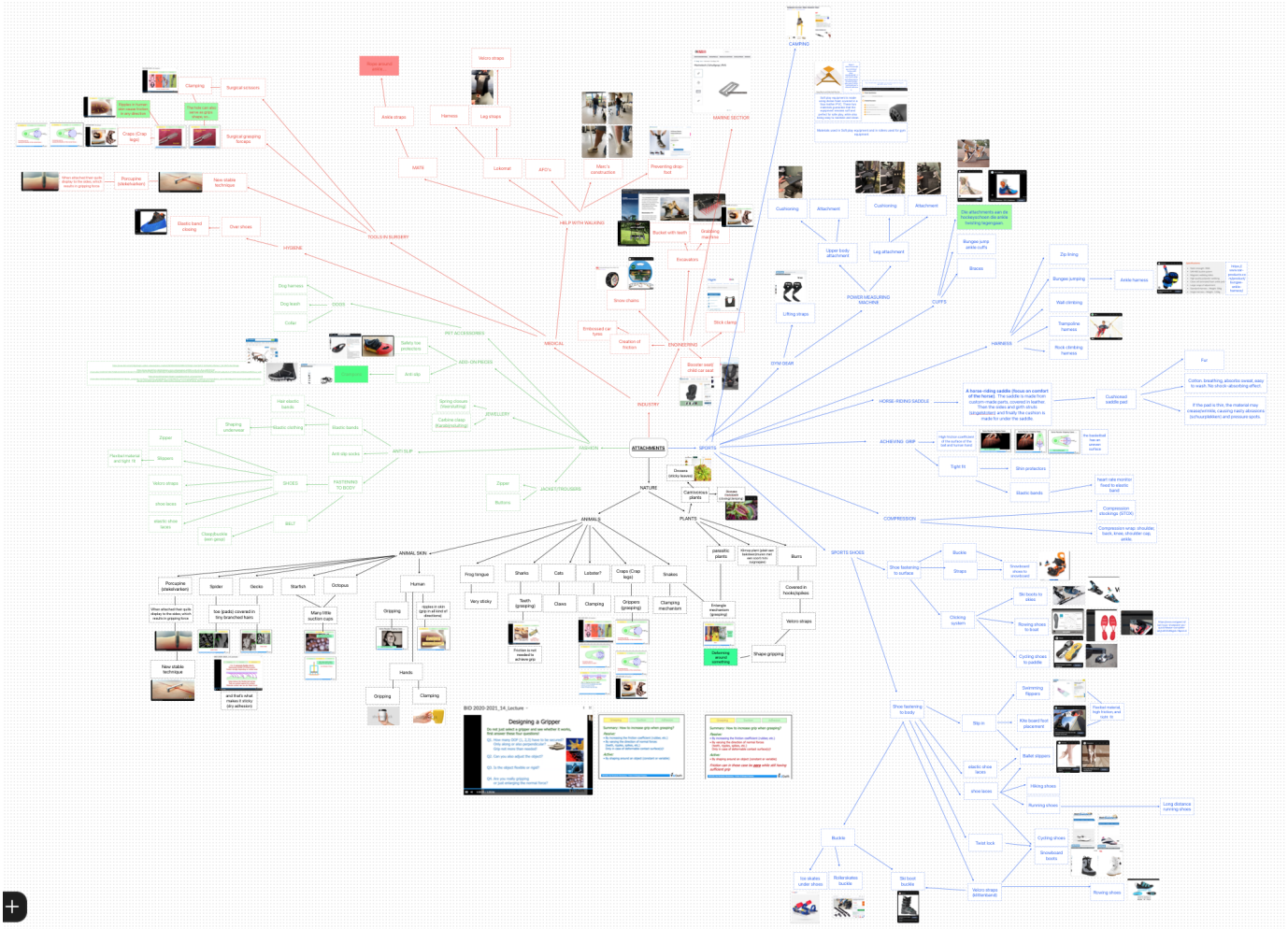


Fig. 12: A mind map is used in this design process to explore different solutions and possibilities for connections and attachments. The mind map displays different fields *(fashion, industry, sports, and nature)* in which a variation of connections and inventive closing solutions have emerged, all solving different types of problems in a similar yet specified way.

3) Ideas, Sketches and Prototypes: After analysing all the information, considering and playing with combinations, sketching ideas, and brainstorming on incorporating all the required features in one design, two concepts were selected. The decision was made not to create attachments with an embedded AFO for two main reasons. First, this would require additional medical requirements. Second, AFOs are specific to each individual. Additionally, it's important to prepare and assist people in returning to their normal lives, and for that, they need to practice with their own equipment.

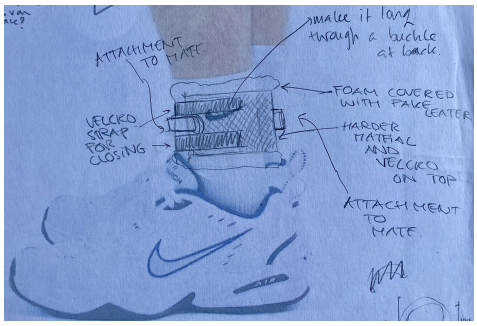
Concept idea cuff design: Inspiration for the cuff design was mainly taken from gym equipment, existing braces (for the arm or leg and the dropfoot brace) and children's toys. All these objects have in common that material wrapped in a comfortable fabric is tied around the body and closed with Velcro. Furthermore, the design was inspired by my experience with a strengthening test device at a physiotherapist in Amsterdam. During intense and powerful movements of the legs, the padded straps covering the leg proved to be comfortable and stayed in place during explosive movements. The outside of the padding was synthetic leather.

The design concept consists of a padded cuff around the leg just above the ankle, making use of the tapered shape of the leg. A C-shaped aluminium plate is surrounded by a thick layer of foam. These two materials are attached to each other with smooth buttons. When the padding is thick enough and the buttons strategically positioned, the buttons will not touch the skin while the MATE is in use. The C-shape cuff is sealed by a padded plate and closed by Velcro straps. A sketch is shown in figure 13a.

Concept idea bandage design: The bandage design is based on and inspired by techniques from different fields. The first inspiration came from hiking crampons, which are devices attached to shoes to improve mobility on slippery surfaces (figure 13b). The wrapping technique around the shoe is sturdy enough to stay in place while walking on these slippery surfaces. It also offers the potential for attaching the MATE's elastic cables in the front and back of the foot. The second source of inspiration came from a wrapping technique used during a gait rehabilitation exercise at the visited rehabilitation centre. During

the gait training of a woman with hemi-paralysis, a bandage was attached to the affected leg by wrapping it around the front arch of the foot, as shown in figure 13c. The physiotherapist was holding the end of the bandage string in his hands. When walking side by side, he could control her affected leg and help her initiate the swing phase by lifting her foot. As the goal of the exercise is similar to the MATE's purpose, this wrapping technique is at the core of the design process.

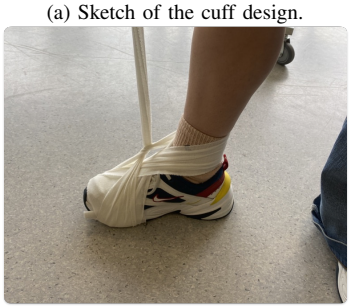
The bandage design combines the two techniques, where two straps wrap around the foot and the ankle and are closed with Velcro; see sketch 13d. As can be seen, one strap wraps around the ankle, and the other one crosses under the foot, creating a sling shape. Only thin straps are placed under the foot to prevent any negative impact on the gait cycle.



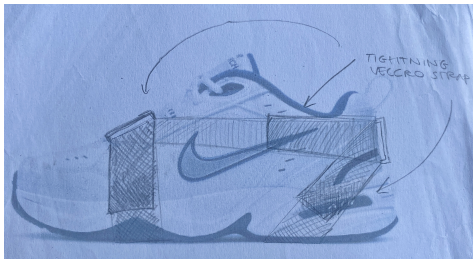
(a) Sketch of the cuff design.



(b) The design of the hiking crampon inspired the bandage design (photo [36])



(c) The complete leg can be operated with only the wrapped bandage design around the anterior arch of the foot. This operation technique is an inspiration for bandage design.



(d) Sketch of the bandage design

Fig. 13: Inspiration and sketches of the attachment designs.

4) Iterations towards final design: In the initial cuff design, the foam padding and synthetic leather covering are affixed to the aluminium C-shaped frame. The Velcro closures are situated on the exterior of the cuff and feature a closure mechanism akin to that of Velcro shoes, where the straps are looped through a ring and fastened back onto themselves.

The initial bandage model comprises solely Velcro straps and a 3D-printed PLA buckle positioned at the front and back of the design. When viewed from the side, the two Velcro straps form a triangular shape, with the flat side of the triangle encircling the ankles (upper strap). The lower strap crosses underneath the foot to the opposite side. Additionally, synthetic leather has been utilized to cover the Velcro upper strap, providing protection to the wearer's skin.

Interview with orthopaedic device maker: Upon completing the initial prototype, we held the meeting (mentioned above) with the orthopaedic device maker to gather feedback. During the session, we presented the synthetic leather and foam materials used in the prototypes for evaluation. It was suggested that the leather side of the material be used 'outwards' for easy cleaning. Additionally, we were advised to prevent wrinkles in the design to avoid trapping dirt. Lastly, we discussed the concept of having a changeable cover around the foam. [34].

Further, when securing the top of an AFO to the leg, the closure can be circular or threaded through a rectangular ring, similar to the closure on Velcro shoes. Circular closure is preferred when the patient has limited hand function or can only use one hand, as it is easier for the patient to manage [34].

Second round of prototypes: After conducting an hour-long walking test with the MATE, a second set of prototypes was developed. In this iteration of the cuff design, the holes in the aluminium C-shaped connectors have been repositioned to allow more space for the padding, preventing it from being overly flattened. Furthermore, the synthetic leather is now loosely wrapped around the design instead of being fixed to the foam and C-shape. Additionally, the closure method has been updated – the closing straps are now wrapped around the cuff for securing, and the 'fold-over' principle is utilised for added strength. This involves the Velcro straps connecting at the back of the cuff by overlapping with each other.

Based on the feedback from the testing round, it is evident that improvements are needed in the quality of the bandage prototype, particularly in which areas require Velcro and which do not. Additionally, it is suggested to incorporate foam between

the upper strap and the wearer's ankle. The decision to switch to a metal buckle is based on the visible damage observed on the PLA 3D printed 'hooks' after the one-hour test walk and concerns about the strength of the 3D printed 'D-ring' to withstand the forces exerted by the MATE on the attachment designs.

In order to develop a new buckle, it was essential to first optimise its dimensions. Consequently, a new size PLA buckle was integrated into the bandage design for the second test phase. The new design for the 3D-printed PLA buckles features larger openings on the side for easier passage of Velcro straps and a smaller ring/opening to attach the hooks. The harness construction (straps) from the previous iteration has been retained. Furthermore, the foam was integrated between the 'upper' straps and ankles, surrounded by synthetic leather.

Testing second round of prototypes & needed changes for the final designs: During the second round of testing, the cuff design failed and detached after 40 minutes of walking due to a production malfunction. However, this issue can be easily and immediately resolved. The sudden release of the design did not cause any pain or problems, and the user was able to continue walking without the design.

Testing the prototypes revealed that for the final cuff design, the synthetic leather should be pulled tightly around the foam and cuff to avoid creating wrinkles. This can be achieved by using Velcro, which would also create a 'double-lock' that provides extra security for the Velcro straps. Furthermore, for the bandage model, a larger piece of foam is required between the upper strap and the ankles to prevent it from slipping away during walking.

Third round of cuff prototype: Before finalising the cuff design, a final test was conducted to optimise it. A cuff prototype with an opening on the side was created. However, this arrangement made it difficult to align the hooks that attach the design to the elastic cables of the MATE in a straight line at the front and back of the design. In the side-opening design, it is crucial to find the perfect size to ensure that the forces are transferred correctly to the user's legs and to keep shear stress minimal. This requires creating many more different sizes of the design. As a result, this design idea was ultimately dismissed.

5) Materials of the designs: During the interview with the orthopaedic device maker, some material properties of the AFO were discussed. The straps of an AFO are made of canvas or leather, usually with a Velcro fastener. Leather and synthetic leather can both be used. Sometimes plastic is used for the straps, and these are easier to clean than the leather and canvas versions [34].

All the materials used for the attachment designs are readily available and affordable. The most prominent materials for the designs are mentioned below.

- *Aluminium cuff and front plate:* A 1 mm thick aluminium plate (aluminium 5754, condition H12) has been used as a base for the cuff to maintain its shape. This plate is easy to bend into the preferred C-shape while staying firm and resilient when used.
- *Foam:* A 3 cm thick foam (cold foam) has been used as padding in both designs to increase comfort.
- *Synthetic leather covers:* The foam enhances comfort, but it should not directly touch the wearer's skin because it can harbour bacteria. That's why synthetic leather (polyester) is used to cover the foam. Synthetic leather is soft for the skin and is easy to clean. Also, it has demonstrated its effectiveness for widespread use, as it is also utilised in gym equipment that comes in contact with people's skin.
- *Lashing straps:* Lashing straps are utilised in the harness construction of the bandage model due to their lightweight, flat, easily moldable but sturdy, and durable qualities.
- *Customised buckles for the bandage model:* The shape of the hook-buckle combination for the bandage model was not available in stores, so it had to be customised. Stainless steel was chosen because this material is commonly used in the medical field.
- *Velcro closing straps:* Velcro straps are regularly used for children's toys and shoes as a closing principle. It's an easy way to attach something as it does not include tying or knotting things or needing high muscle strength. Compared to other fastening options, Velcro also poses a lower risk of over-tightening. In addition to its use in children's items and shoes, Velcro is also utilised in sports equipment, including items with and without weights, as well as in the securing of braces and in the Lokomat.

D. Deliver

The final design concepts are developed and tested in the *Deliver* phase. Based on the experimental results, the designs are analysed and assessed, after which the preferred design is recommended for further development.

1) Final cuff design: The finalised cuff design includes a C-shaped cuff with a diameter of around 10 cm, which can be easily moulded to fit the leg thickness. It is secured just above the ankle. The design comprises interchangeable parts (if similar sized), which are easily replaceable when needed. Each cuff weighs 194 grams.

C-shaped back part: In Figure 14, both the front and back of the cuff attachment used in the experiments are depicted. The illustration highlights the four closing flaps of the cover surrounding the padded C-shaped back part. Beneath this cover, the aluminium C-ring with interior foam and exterior black Velcro is positioned.

Front part: The aluminium rectangular front part is padded with foam on the inside and has black Velcro on the outside. The foam is covered with synthetic leather. Different size covers are designed for the small, medium, and large front parts.

Velcro closings: The two Velcro closing straps that hold the two parts together are equipped with double-sided Velcro. The black soft side connects to the outside of the C-shaped back part (which is also black Velcro), while the white hard side of the Velcro straps points outward. The flaps of the back cover have Velcro inside that connect to both the Velcro on the C-shaped back and the white Velcro straps, creating a double-lock for the Velcro closings. It is recommended that the Velcro on the flaps be white in the next iterations to provide clarity on which flap closes where. Additionally, the flaps of the cover connect to each other with Velcro.

Hooks: A hook fixed to the front connects the design to the front cables, and another hook at the back connects the design to the back cables.

User safety: Both the front section and the C-shaped back are entirely covered with Velcro to maximise contact surface with the Velcro closing straps and back cover flaps. According to force calculations in Appendix B, a connection of 10.2 cm Velcro per strap end is required for safe force transmission in standard conditions. The closing straps measure 42 cm, with at least 16 cm per strap end making contact with the Velcro on the C-shaped back cover.



(a) Front cuff design.



(b) Back cuff design.

Fig. 14: In this figure, the cuff design used in the experiments is shown from both the front and back.

Material per cuff design:

- 1 mm thick aluminium front (aluminium 5754, condition H12)
  * Small: 25 mm x 60 mm; see figure 15
  * Medium: 35 mm x 60 mm; see figure 15
  * Large: 45 mm x 60 mm; see figure 15
- 1 mm thick aluminium C-shape. Aluminium 5754, condition H12. Dimensions plate 270 mm x 60 mm; see figure 16. Diameter C-shape: around 100 mm, but easily moulded to fit the leg thickness.
- Velcro on front (tesa, extra strong)
  * Small: 25 mm x 60 mm, but from the midpoint of the hooks, bolts, and eye studs, leave a 10 mm open space.
  * Medium: 35 mm x 60 mm, but from the midpoint of the hooks, bolts, and eye studs, leave a 10 mm open space.
  * Large: 45 mm x 60 mm, but from the midpoint of the hooks, bolts, and eye studs, leave a 10 mm open space.
- Velcro on C-shape (tesa, extra strong): 268 mm x 50 mm, but from the midpoint of the hooks, bolts, and eye studs, leave a 10 mm open space.
- 2x Velcro closings (Panduro & tesa extra strong): 420 mm x 20 mm
- Foam inside front part. Type: cold foam.
  * Small: 50 mm x 100 mm x 30 mm
  * Medium: 60 mm x 100 mm x 30 mm
  * Large: 70 mm x 100 mm x 30 mm
- Foam inside C-shape. Type: cold foam. Dimensions: 285 mm x 100 mm x 30 mm.
- Synthetic leather (polyester) cover front part
  * Small: 130 mm x 170 mm, with an elastic band of 150 mm.
  * Medium: 140 mm x 170 mm, with an elastic band of 150 mm.
  * Large: 150 mm x 170 mm, with an elastic band of 150 mm.

- Synthetic leather (polyester) cover C-shape, with tesa extra strong Velcro closings.
- 8 eye studs (4 mm wide)
- Nuts M3 (4 per cuff)
- Bolts M3 and rings (4 per cuff)
- 2x attachment hook (steel gamma artnr. 591733)

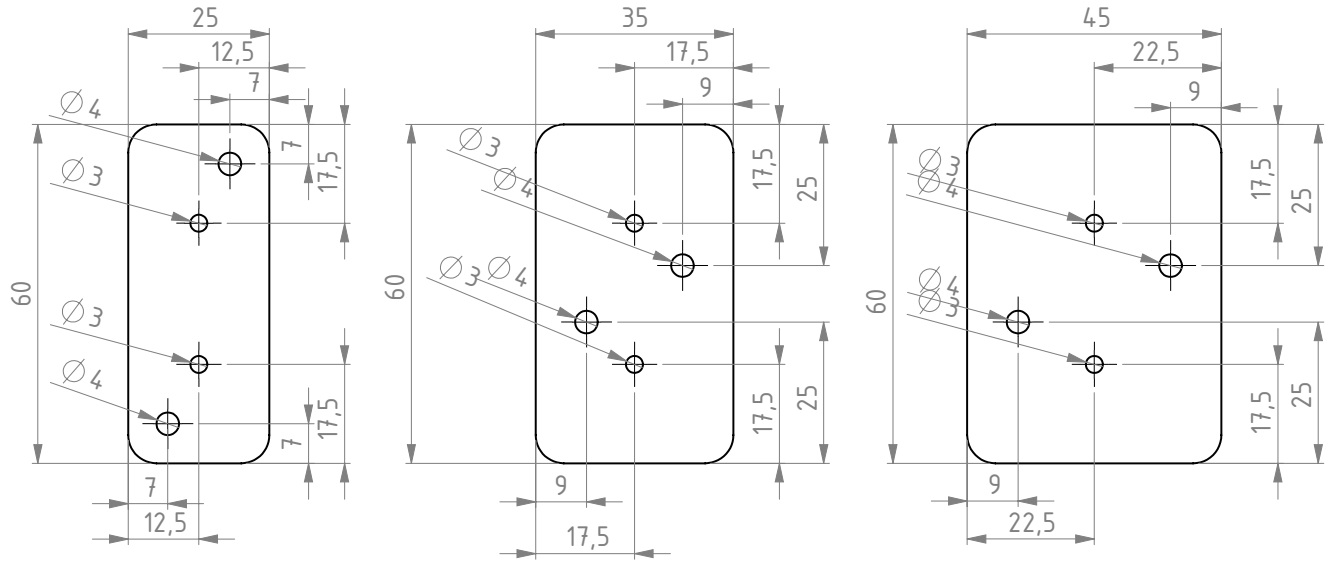


Fig. 15: Aluminium front part from SolidWorks with dimensions (small, medium, large, respectively).

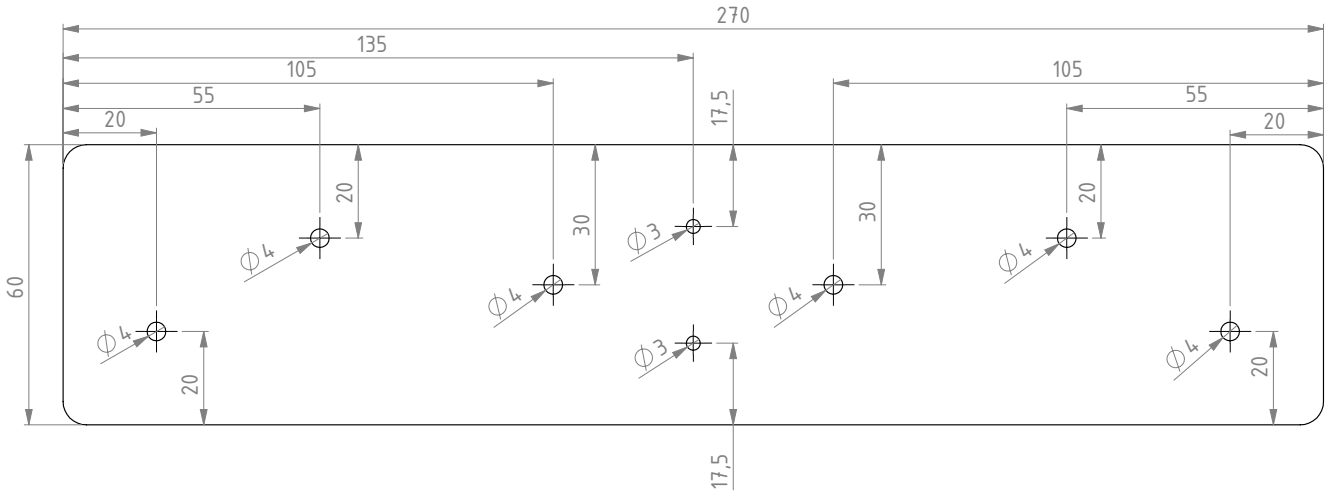


Fig. 16: Aluminium back part from SolidWorks with dimensions.

2) Final bandage design: The final bandage design incorporates an upper and lower strap, as well as two custom hook-buckles and two protective pads. The design securely wraps around the foot and ankle, as illustrated in figure 17b. Each component can be individually replaced, enabling a tailored fit. Moreover, in the event of damage, only the affected part can be easily swapped out. The design weighs 107 grams.

Upper strap: In figure 17a, the green upper strap is depicted. The strap's midpoint is positioned at the back of the shoe, where the strap is wrapped around the middle bar of the posterior hook-buckle. Both ends then encircle the ankle to the front of the leg, wrap around the anterior hook-buckle's outer bars, and are secured in place using Velcro.

Lower strap: In figure 17a, the orange strap denotes the lower strap. The middle of this strap is wrapped around the middle bar of the anterior hook-buckle. Both ends of the strap are then guided under the shoe, crossing and emerging again on the other side of the shoe, closer to the heel, creating an x-shape under the shoe. Then, the ends of the strap are guided back up to each reach an outer bar of the posterior hook-buckle, around which they wrap and fasten themselves with Velcro.

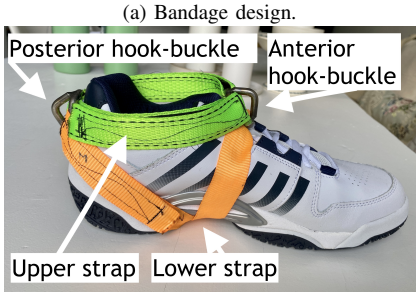
Customised hook-buckle: The customised hook-buckle features four connection points. On the flat part of the buckle, three bars surrounded by green or orange colours serve as connection points for the straps. At the anterior hook-buckle, the middle bar is orange, and the outer ones are green. Meanwhile, at the posterior hook-buckle, the middle one is green, and the outer ones are orange. The hook part stands perpendicular to the flat buckle and acts as a connection point from the attachment to the elastic cables of the MATE. The shape of the hook-buckle is depicted in figure 18. The paper can also make reference to this hook-buckle as a buckle, clasp or hook of the bandage design.

Protection pads: A protective pad is positioned between the upper strap and the wearer on both sides of the ankle. Each pad consists of a layer of foam surrounded by a synthetic leather cover. Three types of pads have been created to determine the most comfortable and effective option, see figure 19. In all versions, the upper strap falls into the appropriate gap and is then 'locked' so the pad is secured in place during walking. All pads have the same dimensions.

User safety: The straps are equipped with Velcro on certain sections. According to the force calculations in appendix B, a minimum of 14.3 cm of Velcro is required at both ends of the upper strap and 9.6 cm at both ends of the lower strap to ensure safe force transmission in a normal situation. The lower strap fulfils this requirement; even the smallest version has 11 cm of Velcro on the lower strap ends. The smallest upper strap has 14 cm of Velcro on each end. During prototype testing, this Velcro length proved to be safe.



(a) Bandage design.

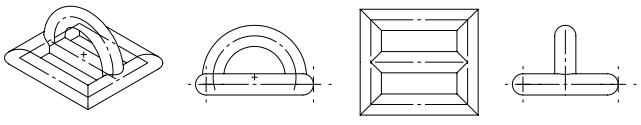


(b) Bandage design around a shoe, without protection pads.

Fig. 17: In this figure, the bandage design used in the experiments is shown in different conditions but without the protection pads. The experimental set-up of the bandage design with protection pads is shown in figure 2c.

Material per bandage design:

- 2x customised austenitic RVS hook-buckles. Material: P+P TP 316 L ASTM-A 269 SEAMLESS 6:35 x ..., 6 mm diameter. Dimensions: 30 mm x 38 mm x 22 mm. Colours: orange and green.
- Lashing straps bright orange: Carpoint Spanbanden, breaking strength: 500 kg (Gamma)
- Lashing straps bright green: Jumbo LC125daN (Gamma)
- Velcro straps: Panduro
- Foam cushions on the ankles surrounded by synthetic leather (polyester): 100 mm x 80 mm x 30 mm. Type foam: cold foam.
- Upper strap (lashing strap)
  * Medium: 67 cm x 2.5 cm, with 14 cm x 2 cm Velcro per strap end
  * Large: 70 cm x 2.5 cm, with 15 x 2 cm cm Velcro per strap end
- Lower strap (lashing strap)
  * Small: 90 cm x 2.5 cm, with 11 x 2 cm cm Velcro per strap end
  * Medium: 94 cm x 2.5 cm, with 11 x 2 cm cm Velcro per strap end
  * Large: 100 cm x 2.5 cm, with 12 x 2 cm cm Velcro per strap end



(a) Hook-buckle design from Solidworks.



(b) Anterior hook-buckle design with coloured bars.

Fig. 18: The customised hook-buckle design from SolidWorks to reality.

3) Detailed instructions of design connection to cables: **Carabiners:** The ends of the high and low cables on the right side are looped through a stainless steel carabiner (GAMMA product nr. 432712). This loop is securely fastened by two metal plates pressed forcefully together, with a knot at the end of the cables for additional security, as depicted in figure 20. The right-side high and low cable ends are looped through a carabiner together and, via this carabiner, connected to a force sensor, which in turn connects to the right attachment design.

The left end of each cable is attached to a cable tension hook (see figure 21). The cable tension hooks in the back are connected to a stainless steel carabiner (GAMMA product nr. 432713), which in turn connects to the back of the left attachment design. The same process is followed for the cable tension hooks in the front, which are connected to the front of the attachment design.

The carabiners used can safely sustain forces up to 110 kg and 130 kg, respectively.

Cable hooks that allow for adjusting the cable length: In addition to the carabiners used to connect the cables to the attachment designs, four cable tension hooks, see figure 21, are utilised to offer a discreet way to adjust the cable lengths if needed. These cable tension hooks will be fastened to the participant's left foot, away from the ZED-mini camera and the force sensors located on the right side of the body. The tension hooks are purchased from Nite Ize and have a weight capacity of up to 227 kg [37].

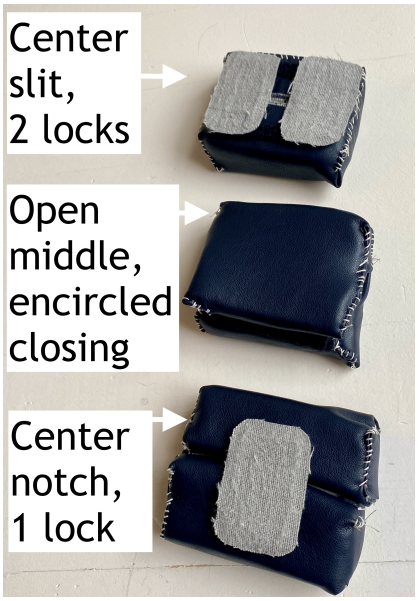


Fig. 19: Types of protection pads.

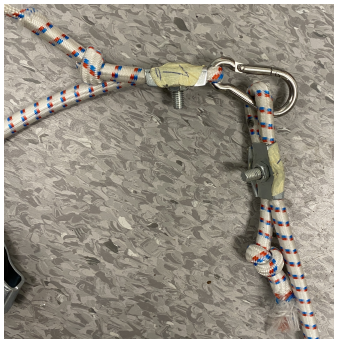


Fig. 20: The right-side high and low cable ends are looped through a carabiner. This carabiner is connected to a force sensor attached to the right attachment design.

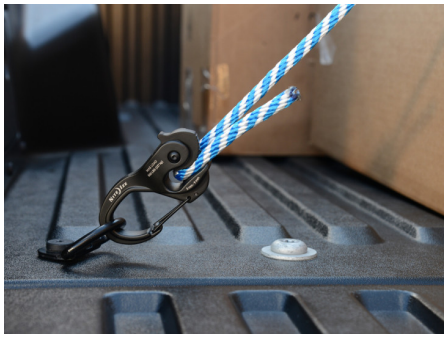


Fig. 21: Four CAMJAM XT cable tension hooks are used to adjust the lengths of the elastic cables when necessary. The photo is from NiteIze [37].

APPENDIX B
FORCE CALCULATIONS ON ATTACHMENT DESIGNS

A. Force calculations & Velcro strap length

The force calculations analyse two scenarios involving a snapshot of an unimpaired person's walking pattern: normal walking scenarios 1 and 2. The calculations reveal that the forces of the cables on the attachment designs and, consequently, on the wearer's legs are consistently highest in the frontal plane (x-direction) (see column 3 in Table IV). Additionally, the force calculations assess the required Velcro length to safely transfer the MATE's forces to humans. Velcro length per strap was determined for normal situations, as well as worst-case scenarios where some straps might detach, and all forces would be transferred to the remaining straps. See chapters 5.1 and 5.2 in the force calculations document in Appendix I for more details. The safety factor used in the calculations is 1.

The minimum length for each Velcro strap for the cuff straps during unimpaired walking is 10.2 cm. For the bandage design, the minimum length is 14.3 cm for the upper strap and 9.6 cm for the lower strap. In the rightmost two columns of Table IV, you can find the Velcro lengths for scenarios where one of the straps detaches.

The force calculation document presents scenarios in which the user's legs are positioned at the back or front of the treadmill, as well as a stiff leg scenario. However, as the MATE set-up involves unimpaired participants using a body weight support system to stay centred on the treadmill and a stop button for immediate intervention, only the results of normal walking scenarios were considered in creating the attachment designs. The force calculation document can be found in Appendix I.

TABLE IV: This table shows the x, y, and z forces acting on the designs' anterior and posterior during normal walking. The corresponding requirements for the Velcro strap length are also provided.

Scenarios		Pulling forces on the legs in x, y, and z-directions [x y z] (in N)	Magnitude pulling forces (in N)	Velcro length per cuff strap	Velcro length per bandage strap	Velcro length per cuff strap (worst-case)	Velcro length per bandage strap (worst-case)
Normal walking 1	Right foot: heel strike	Anterior: 275.65 -77.02 99.89 Posterior: -213.23 -23.40 32.45	Ant.: 303.14 Post.: 216.95	9.6 cm	Upper str: 12.1 cm Lower str: 9.6 cm	19.1 cm	Upper str: 21.7 cm Lower str: 33.7 cm
	Left foot: toe-off	Anterior: 321.34 36.99 38.53 Posterior: -187.21 48.05 49.02	Ant.: 325.75 Post.: 199.40	10.2 cm	Upper str: 14.3 cm Lower str: 5.4 cm	20.4 cm	Upper str: 18.7 cm Lower str: 32.6 cm
Normal walking 2	Right foot: mid-stance	Anterior: 300.04 -50.55 65.00 Posterior: -199.52 -31.85 41.31	Ant.: 311.14 Post.: 206.23	9.6 cm	Upper str: 13.2 cm Lower str: 6.3 cm	19.1 cm	Upper str: 19.4 cm Lower str: 32.5 cm
	Left foot: mid-swing	Anterior: 303.74 56.32 33.54 Posterior: -201.84 35.55 20.51	Ant.: 310.73 Post.: 205.97	10 cm	Upper str: 14.2 cm Lower str: 5.2 cm	19.9 cm	Upper str: 20.3 cm Lower str: 33.5 cm

B. Comments after data analysis experiment

In the data analysis of the experiments, the average pulling forces on the front and back of the right design were obtained. The results indicate that the average maximum force at the front of the designs does not exceed 141 N, and at the back of the designs, it does not surpass 115 N, see Table I. The highest force exerted on the front of the designs across all participants was 190 N, while the corresponding force exerted on the back of the design at that time was 32 N. Additionally, the averages of the highest force of every participant per gait session were calculated, resulting in an average of 119 Newton on the cuff's anterior, 109 Newton on the cuff's posterior, 146 Newton on the bandage's anterior, and 110 Newton on the bandage's posterior (Figure 25).

The original force calculations were based on the most extreme situations when considering assumptions. The forces measured in the actual experiments are remarkably lower than those obtained and used in the calculations. For subsequent design iterations and follow-up research on the attachment designs, getting more precise data on the tension applied to the design straps is important. If they can be applied properly, strain gauges could be used for this.

APPENDIX C
CASE STUDIES

A. Case Study One

Participant 8 fell within the first minute of the experiment when walking with the bandage model. A case study was conducted to better understand what happened. Figure 22 illustrates the forces and kinematic data of the participant during the fall, as well as the first minute of the baseline data and the data of the second trial of the gait session with the bandage model, where the front high cable was lengthened, and the treadmill speed was slowed down from 3.5 km/h to 2.5 km/h.

Before the fall, the video captured by the ZED mini shows that the participant's legs were considerably pulled forward at the toe-off. The participant couldn't make a heel strike, and the toe-off was shortened. Just before the fall, the participant started taking longer steps, and at the moment of the fall, their right leg was pulled forward and upwards too hard. As a result, at the moment of toe-off of the left foot, the right leg was too unstable and placed too far forward to facilitate a good toe-off. The left foot moved further back than normal on the treadmill, resulting in a higher pulling force on the left ankle at the moment of toe-off. The same movement occurred in the next step, leading to the participant losing control over their legs and falling. The treadmill was immediately stopped. The body weight support system caught the participant during the fall, and the participant reported no physical issues afterwards. The bandage design remained securely around the ankle despite the high forces.

The moments before and during the fall, as captured by the ZED-mini camera, aligned with the force and kinematic information in the graphs. In the middle column of figure 22, the moment of the fall is clearly visible. The peak force occurred just before the step height increased at 36 seconds. Subsequently, it can be observed that the ankle moved further back with each step, and the step height remained higher than before. The force graph during this time frame also shows the high forces on the right ankle.

In the first gait session with the bandage design just before the fall, the force at the front of the design peaked at 235.6 N, with a counterforce at the back of 49.1 N. During the 20-minute gait session of trial 2, where the front high cable was elongated, and the walking speed was reduced to 2.5 km/h, the highest force measured at the front of the design was only 171.5 N, with a 43.1 N counterforce at the back. The change in force at the front of the design is clearly visible in the force graphs of figure 22. The highest peak force at the back of the design increased during trial 2, from 99.1 N during the first session to 109.9 N in the second trial, with 72.0 N and 56.0 N, respectively, as counterforces at that time. This can also be seen in table V.

Table VI provides the pulling forces at the front and back of the right attachment design when standing still in the middle of the treadmill, obtained from both trials and conditions.

TABLE V: The max. peak forces and the matching counter-force during the gait sessions of Participant 8, with the bandage design, are presented in the table. In figure 22, the max. peaks during the fall are indicated in the graph. The triangles represent the max. peaks of the front and back force, and the asterisks the matching counter-forces.

Pulling Force (in N)	P8 fall Bandage design	P8 Bandage design
Max. force front	235,6	171,45
Force back when max. force front	49,06	43,07
Max. force back	99,1	109,89
Force front when max. force back	71,97	57,99

TABLE VI: The pretension in the front and back of the right attachment design of both trials of Participant 8, obtained during a standstill in the middle of the treadmill (columns 3 and 4). The lengths of the elastic cables causing the pretension are also shown in columns 5 and 6. No tension was exerted on the cables while they were being measured.

Participant nr.	Front or back force sensor	Average pretension: Cuff design (in N)	Average pretension: Bandage design (in N)	Cable high (in m)	Cable low (in m)
Case study	Front	-	149,10	0,69	1,29
P8 fall	Back	-	62,20	0,97	1,69
P8	Front	118,62	119,01	0,81	1,29
	Back	59,99	60,25	0,97	1,69

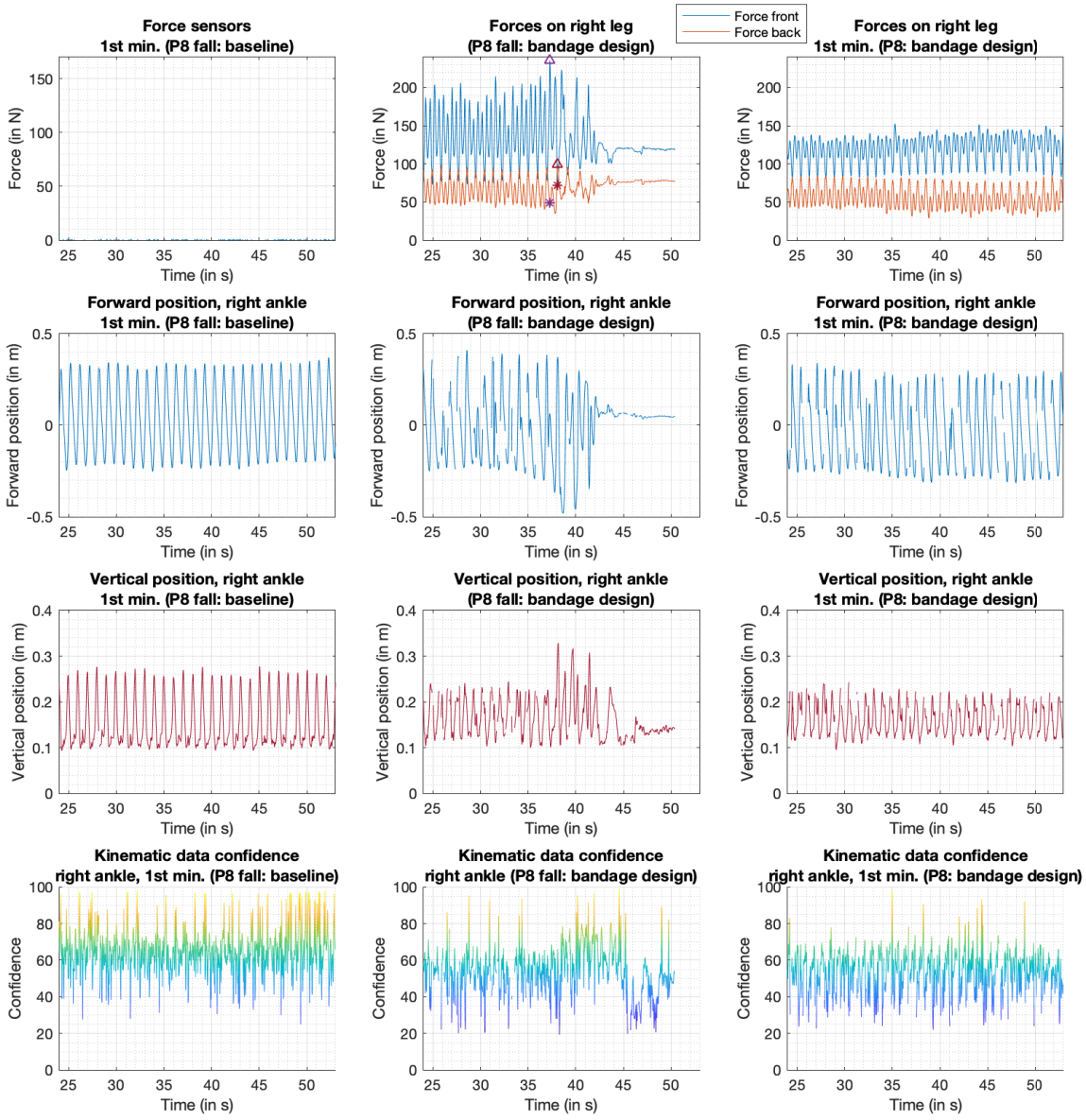


Fig. 22: In this figure, the baseline kinematics of Participant 8 are shown next to the kinematics of the gait sessions with the bandage design, together with the applied forces. In the middle column, the information during the fall is presented. The other columns match the timings to show what happens around that time during the baseline and the additional gait session with the bandage design. The values corresponding with the max. peaks (illustrated with triangles) are noted in table V.

B. Case Study Two

Following the experiments, the stroke rehabilitation physiotherapist previously involved in the interview during the 'Discover' phase was asked to evaluate the MATE using both designs. The physiotherapist conducted a shortened version of the experiment, with 10-minute walking sessions instead of 20 and a 1-minute baseline instead of 5 minutes. This case study presents the physiotherapist's feedback and experiences with the MATE and the attachment designs. The table below (Table VII) contains the cable lengths and corresponding pretensions on the front and back of the designs.

TABLE VII: The pretensions in the front and back of the right attachment design during the first and second trials of the case study with the physiotherapist are shown in columns 3 and 4. The information is obtained during a standstill in the middle of the treadmill. In columns 5 and 6, the cable lengths during each trial are presented. No tension was exerted on the cables while they were being measured.

Case study	Front or back force sensor	Average pretension: Cuff design (in N)	Average pretension: Bandage design (in N)	Cable high (in m)	Cable low (in m)
Physio: trial 1	Front	129.58	123.16	0.76	1.36
	Back	54.17	61.16	0.97	1.69
Physio: trial 2	Front	-	120.15	0.79	1.40
	Back	-	84.15	0.97	1.70

1) Results short experiment: During the short experiment (trial 1), it was observed that although the cable length at the cuff and bandage sessions is the same, the force strengths differ. The mean pulling forces are close, with 98.2 N on the front at the bandage design and 96.4 N on the front at the cuff, and 74.8 N and 81.2 N, respectively, at the backs. The main difference between the two designs lies in the variation of maximum and minimum force peaks. The bandage model shows greater fluctuations between the extremes compared to the cuff model, as seen in the force peaks in the bandage condition in figure 24. The mean maxima and minima of the pulling forces on the designs can be found in table VIII.

The average step lengths of both design conditions are only one centimetre apart and shorter than the baseline step length. When walking with the designs, there is an earlier start of the toe-off and a quicker heel strike, visible in figure 23. Furthermore, the data shows that the average step height of the cuff design is slightly higher than in the baseline, while in the bandage condition, the average step height is slightly lower than in the baseline. The average values are provided in table VIII, and the differences in kinematics between baseline and walking with the MATE in the attachment designs can be observed from figures 23 and 24.

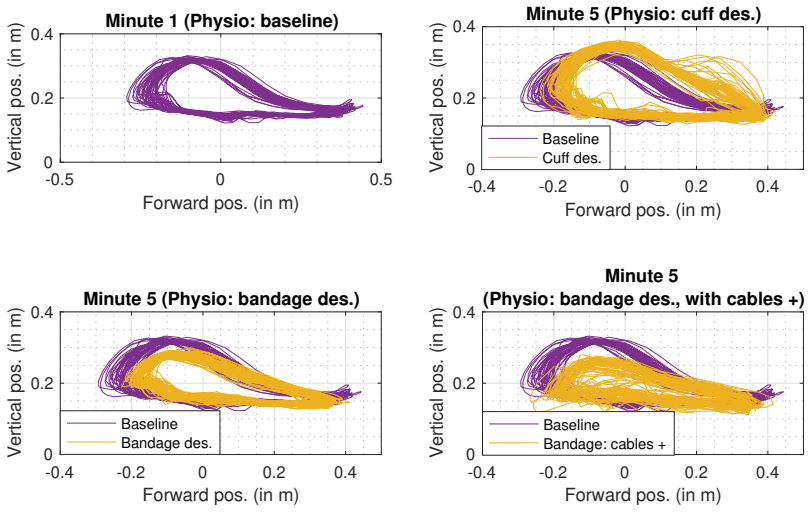


Fig. 23: This figure illustrates one minute in the right ankle trajectories of the physiotherapist in all scenarios. It compares the ankle trajectories during the gait session with the cuff and the two trials with the bandage model to the baseline ankle trajectory.

2) Trial 2: Gait session with the bandage design, including longer front cables: After completing a shortened version of the experiment, the front cables were slightly elongated, and the physiotherapist walked with the bandage model for an additional 10 minutes. Despite the longer front cables, the average forces at the front and back of the design increased, and the fluctuation between the maximum and minimum peaks lessened. Additionally, the average step length decreased; see table VIII and figure 24.

TABLE VIII: This figure presents the average forces on the front and back of the designs, the average step length and the average step height of all conditions during case study two. Additionally, the average step height + snapshot ankle height is noted to provide another perspective on the average step height.

Case study physio	Average step length (in m)	Average step height (in m)	Average step height + snapshot ankle height (in m)	Average pulling force front (in N)			Average pulling force back (in N)		
				Max.	Min.	Total av.	Max.	Min.	Total av.
Baseline	0.615	0.156	0.295	-	-	-	-	-	-
Cuff design	0.551	0.189	0.328	111.2	72.7	96.4	96.9	68.2	81.2
Bandage design	0.541	0.137	0.276	123.4	62.1	98.2	99.6	56.3	74.8
Bandage design (longer front cables)	0.501	0.119	0.283	126.8	75.3	105.9	114.6	76.2	92.3

3) Questionnaire results physiotherapist: The data collected from the participants was also gathered from the physiotherapist and recorded in table IX. This indicates that the physiotherapist experienced a better user experience while walking with the bandage model. However, both models scored slightly above 0 on the -2 to 2 scale. Additionally, the MATE acceptance is notably higher when the bandage model is worn. As for the acceptance of the designs, the scores are equal on the usability scale, while in satisfaction, the bandage model consistently leads. The cuff model excels in design usability, scoring 90 compared to 67.5 on a 0-100 scale.

TABLE IX: The physiotherapist provides the same information as the participants during a shortened experiment version. This table includes the physiotherapist's demographics, design order, treadmill speed, questionnaire responses, donning and doffing times, and differences in cuff height. The questionnaire scales range from -2 to 2 for the user experience and system acceptance questionnaire and from 0 to 100 for the design usability questionnaire. Additionally, the system acceptance questionnaire includes the scales for usefulness (U) and satisfaction (S).

Case Study physiotherapist													
Demographics and general info	Gender		Age (in years)		Height (in m)		Weight (in kg)		First tried attachment		Second tried attachment		Treadmill speed (in km/h)
	Man		49		1.81		73		Cuff		Bandage		3
Questionnaires	User experience		MATE acceptance with Cuff		MATE acceptance with bandage		Acceptance Cuff		Acceptance Bandage		Design Usability		
	Cuff	Bandage	U	S	U	S	U	S	U	S	Cuff	Bandage	
	0.1	0.38	0.6	0	1	1	0.8	0.13	0.8	0.5	90	67.5	
Donning/doffing physio and cuff height	Donning time (in s)		Doffing time (in s)		Cuff height differences (in m)								
	Cuff	Bandage	Cuff	Bandage	After donning	Before 10 min walk, after cable attachment		After 10 minute walk, cables still attached					
	94	169	19	30	0.14	0.15		0.18					

4) Notes and quotes of the physiotherapist on the attachment designs and the MATE, during trials 1 and 2: During the gait sessions with the MATE, the physiotherapist made observations and comments. The comments reflect what is visible in the graphs, as the physiotherapist noted that, e.g., the swing phase starts earlier when walking with the MATE than when walking in the normal gait cycle. Additionally, it appears that the physiotherapist is enthusiastic about the effect of the MATE with both designs but clearly prefers the bandage design. The bandage model was found to transfer forces from the MATE to the user better. The physiotherapist recommends focusing on developing the bandage model only and improving it by ensuring there are no loose protection pads and placing the anterior connection point with the cables lower on the forefoot.

During the initial session with the bandage model, the physiotherapist noted that the front cables were a bit too tight. In the session with the extended cables, the physiotherapist reports that walking with the longer anterior cables is easier and that the MATE still provides support. He also mentioned that his feet are now guided rather than controlled. Additionally, he indicated that walking with the MATE becomes easier with experience. During this gait session, he further noted that toe-off can be completed better, and the start of the swing phase is more controlled and less abrupt, although this is not clearly reflected in the kinematics graphs. The physiotherapist also mentioned that with the extended cables, walking feels more natural and that walking with the bandage model is easier than with the cuff design. He stated, "With the cuff, walking was not energy-saving (you're trying to correct yourself more), and with the bandage, it was always energy-saving. But now that the cables have been adjusted, walking feels more natural. Own gait was interrupted last time (with the shorter cables), but now own gait can be completed much more."

The notes and quotes: The first attachment design the physiotherapist tried on was the cuff design, so the first reactions to walking with the MATE (and the cuff model) are mentioned below.

Notes and quotes on the MATE with the cuff design

- Physio mentions that walking feels unnatural and takes more energy than normal.
- Physio does not think walking with the MATE (and the cuff design) is comfortable but does state, "This is more pleasant than walking with the Lokomat", expressing that with the MATE, he has much more freedom of movement and walks more naturally.
- The Physio talks the whole time throughout the 10-minute walk, so talking and walking simultaneously looks like it is going well for the participant.
- Physio mentions that he gradually adjusts the timings of his gait cycle according to the effect of the MATE on his legs.
- Physio thinks it is very COOL that such a mechanical thing could replace the Lokomat! He immediately mentioned the possibilities for every physiotherapy practice and the possibilities for low-income countries.
- He also mentioned that extra research on the ideal treadmill speed and cable length is not prioritised, as that has to be determined on a patient-by-patient basis.
- Physio said he would like to return with different types of AFOs to see and test how they react to the MATE and current attachment designs.
- Physio suggests that the cuff design may fit around an AFO, depending on the AFO type. He also suggests that patients may experience foot-dragging when wearing an AFO.

- "The start of the swing phase is happening more quickly than normal."
- "Tension in the cables higher at the front."
- "When you walk more upright (instead of hunched over a little), your feet are pulled out from under you a bit."
- "Patients would need to practice this."
- "The MATE is COOL! It is amazing that you can replace a whole robot like that!"

Secondly, the bandage design was put on for another 10-minute gait session with the MATE. The notes and quotes of the physiotherapist's experience are given below.

Notes and quotes on the MATE with the bandage design

- Physio mentioned he is curious about the effect of making the anterior connection point on the bottom of the shoe laces (closer to the toes) instead of the top of the laces (closer to the ankle).
- Physio's feet are lifted when walking with the MATE, but not so much that it is enough for someone with a drop foot.
- Physio says he is 'in control' of his feet/body but mentions that if you are taken out of your walking 'flow', it is difficult to get back into it.
- Physio indicates that some AFOs will work with this design, while others will not. For example, the rear splint AFO works, but the toe-off AFO does not.
- What would you change about the attachment designs?
    - o Bandage: An anterior cable connection placed more on the lower front of the foot (see first bullet point). Also, having the protection pads separately is awkward, as you want much more click-and-play. Donning should go from donning to walking on the treadmill in 2 minutes.
    - o Further development should be with the bandage design, not the cuff.

- "It definitely feels different than with the other design."
- "Again, that start of the swing goes quicker than normal."
- "Cables might be a bit too tight in the front."
- "The start of the swing right after toe-off is cut short." (i.e. the first phase of the swing phase)
- "With this design, the connection/power transfer from the MATE to the human is more pleasing." and "It feels more natural to make a swing movement with your foot instead of a movement perpendicular to the treadmill."
- "You can apply your own gait when walking, and you can compensate your movements if needed, whereas, with the Lokomat, you are 'locked'." The physiotherapist would always choose the MATE over the Lokomat because of the freedom of movement.
- "This design worked better because it allows your knee joint to roll and slide back here, keeping it in the same place in the knee. (Rolling is caused by knee extension and sliding back because the knee head must always remain in the same place relative to the knee socket (this is what the knee ligaments are for, among other things).) With the cuff design, your shinbone was pulled forward more horizontally (where the leg stays perpendicular to the treadmill), so the rolling of the knee joint happened, but the sliding back did not because the forces forward were too high and, therefore, pulled too much/hard on your leg."

After the short experiment with the physio, another trial with the bandage design was performed, in which the two cables in the front were elongated; see table VII for the measurements and the tension on the cables during standstill.

Notes and quotes on the MATE with the bandage design with longer front cables

- Physio can comfortably do 2 things at once: walking and talking and walking and looking around.
- Physio comments that walking with MATE gets easier with practice.
- Physio makes only a little heel strike (less than normal).
- Physio indicates that both legs are being pulled equally strongly.
- Physio suggests that for the next research, participants are asked if they have leg problems and that this is added to the same information as gender, age, etc.

- "This walks easier." (The lengths of the front two cables have been adjusted (longer)).
- "MATE does still assist."
- "Toe-ff can be completed better. And the start of the swing phase is more controlled and less abrupt."
- "The feet still get pulled forward and upward. Not so much outwards, but you walk slightly more wide-legged than normal."
- "With the cuff, walking was not energy-saving (you're trying to correct yourself more), and with the bandage, it was always energy-saving. But now that the cables have been adjusted, walking feels more natural. Own gait was interrupted last time (with the shorter cables), but now own gait can be completed much more."
- "Feet are now more guided than controlled."

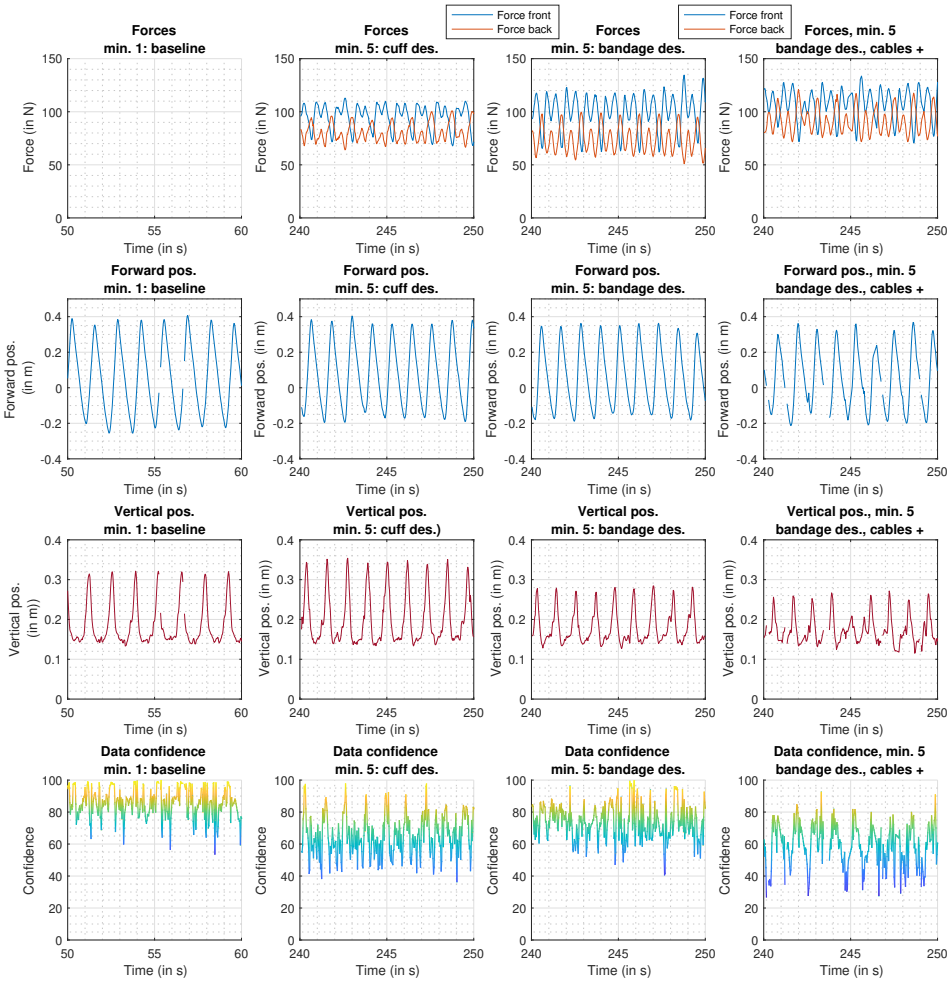


Fig. 24: In this figure, the forces on the attachment designs and the kinematic data of the right ankle from a fixed time period during the gait sessions with the designs and baseline are shown side by side. The first three columns contain information from the abbreviated version of the experiment conducted by the physiotherapist. The rightmost column displays data from an additional trial with the bandage design featuring extended anterior cables. The cable lengths are detailed in table VII.

APPENDIX D
EXPERIMENTAL RESULTS

A. Additional Figures and Tables

TABLE X: Per participant and condition, the pretension in the front and the back of the right attachment design are obtained during a standstill in the middle of the treadmill (columns 3 and 4). The lengths of the elastic cables causing the pretension are also displayed in columns 5 and 6. It is important to note that no tension was applied to the cables during the measurements, so the lengths provided are the unstretched lengths.

Participant nr.	Front or back force sensor	Average pretension: Cuff design (in N)	Average pretension: Bandage design (in N)	Cable high (in m)	Cable low (in m)
P1	Front	126.5	145.4	0.69	1.29
	Back	61.74	58.40	0.97	1.69
P2	Front	131.33	130.8	0.72	1.32
	Back	56.29	55.84	1.00	1.71
P3	Front	114.1	112.13	0.69	1.29
	Back	64.95	58.85	0.97	1.69
P4	Front	143.47	145.34	0.69	1.29
	Back	47.85	58.96	0.97	1.69
P5	Front	141.88	139.46	0.72	1.3
	Back	46.13	54.20	0.97	1.71
P6	Front	112.68	112.79	0.79	1.29
	Back	62.48	58.02	0.97	1.69
P7	Front	119.38	112.01	0.79	1.3
	Back	58.87	68.29	0.97	1.69
P8	Front	118.62	119.01	0.81	1.29
	Back	59.99	60.25	0.97	1.69

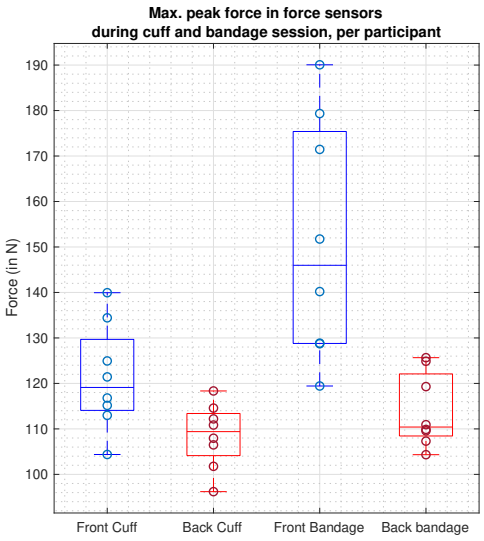


Fig. 25: This box plot shows the maximum peak force in the front and back of the design from the cuff and bandage gait sessions, per participant. Also, the average of these maximum peaks in the front and the back of the design are marked in the figure. The median force of the Front cuff = 119 Newton, Back cuff = 109 Newton, Front bandage = 146 Newton, and Back bandage = 110 Newton.

TABLE XI: This table shows the average forces on the front and the back of the right attachment design. The average of the maximum peaks, minimum peaks, and the total data of the cuff and the bandage design are noted.

Pulling Force (in N)	P1		P2		P3		P4		P5		P6		P7		P8	
	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design	Cuff design	Bandage design
Average max. force front	117.68	138.06	108.41	121.88	101.78	107.20	104.68	125.95	114.92	140.58	104.92	110.75	100.55	108.63	96.22	115.68
Average min. force front	81.06	81.30	65.22	67.72	81.18	80.17	59.43	60.79	63.73	65.43	59.49	53.50	57.08	55.36	63.10	69.88
Average pulling force front	**102.48**	**114.02**	**91.86**	**97.37**	**94.64**	**95.99**	**88.09**	**98.41**	**96.21**	**108.79**	**86.76**	**87.74**	**84.09**	**86.20**	**83.94**	**96.32**
Average max. force back	89.47	94.37	98.16	97.62	110.22	109.49	103.04	110.40	95.53	99.64	101.80	99.90	106.19	114.24	101.82	95.03
Average min. force back	65.23	59.93	69.24	59.57	81.18	67.48	75.45	68.06	63.51	54.31	70.19	58.27	73.03	70.33	76.68	58.34
Average pulling force back	**76.41**	**75.11**	**81.70**	**77.52**	**93.86**	**85.70**	**86.85**	**86.06**	**77.38**	**73.32**	**83.78**	**75.14**	**86.94**	**88.56**	**86.96**	**74.80**

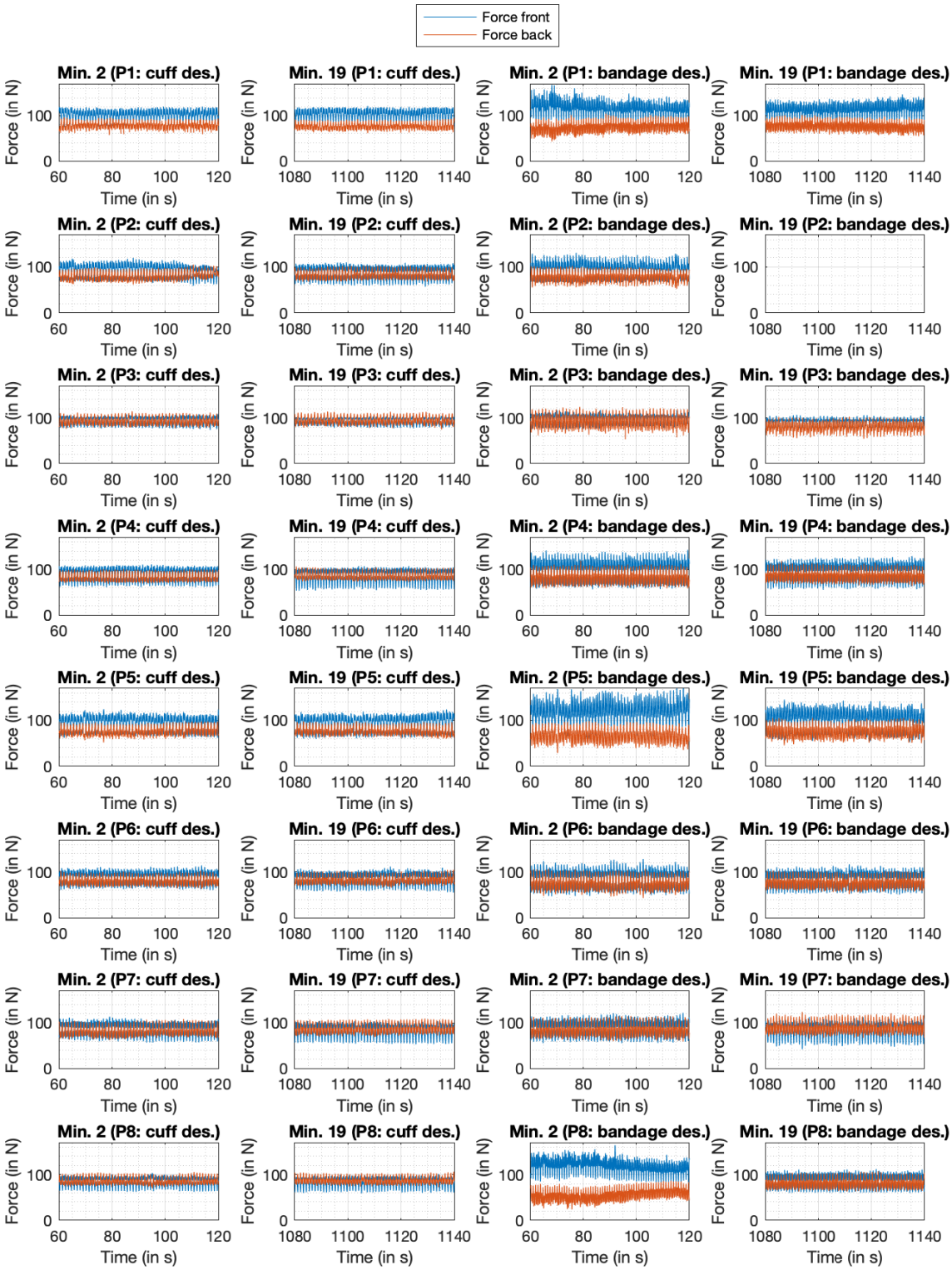


Fig. 26: This figure shows the 2nd and the 19th minutes of the sessions with the cuff and the bandage design. Differences can be observed between the minutes and between the different conditions. (There is no data of the 19th minute of Participant 2 with the bandage design, as the session was stopped before the minute was reached.)

Figure 27 shows the forces and kinematics at different times during a gait session of Participant 8 with the bandage design. The differences in forces and kinematics might indicate there is a link between the changing force magnitude and the changing ankle kinematics of the user. One cannot tell whether the same relationship occurs between the cuff design and the user kinematics, as no data with large force differences between minus 2 and minus 19 is available with the cuff design.

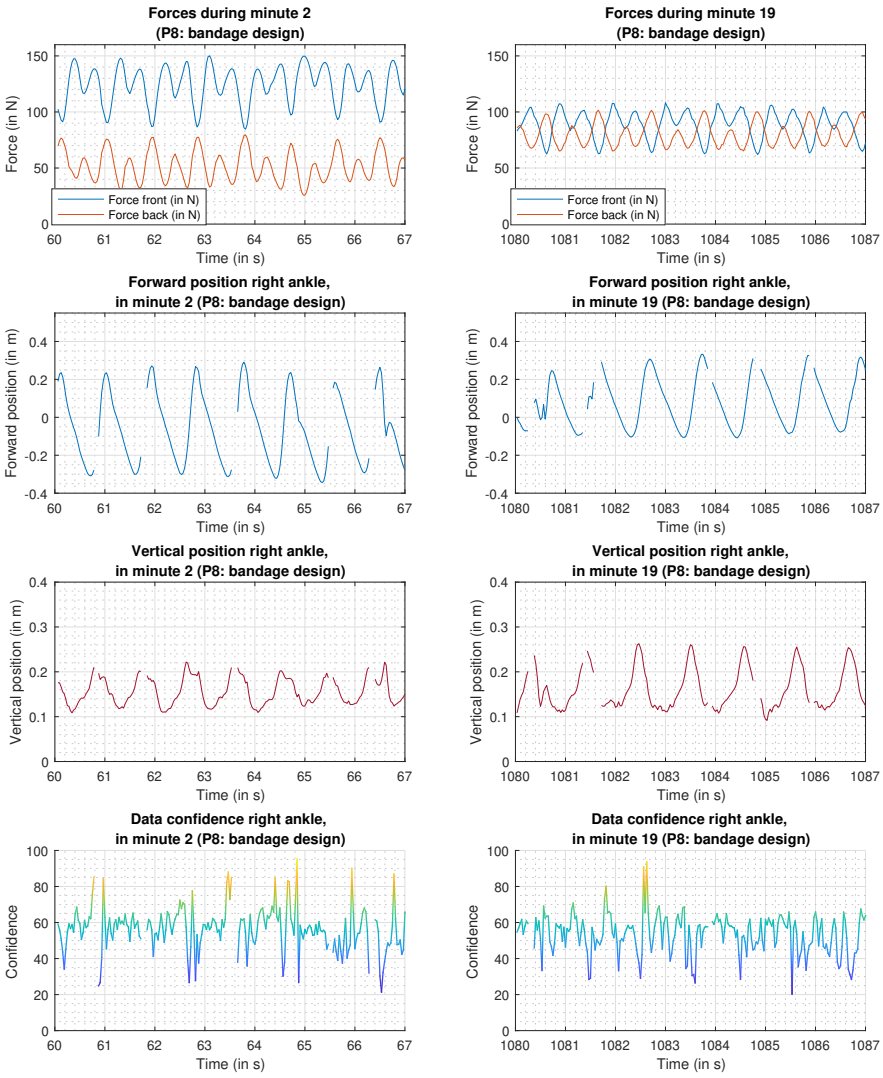


Fig. 27: Kinematics in the forward (frontal plane) and vertical direction (transverse plane) and the forces during the first 7 seconds of the 2nd and 19th minute of Participant 8 walking with the bandage design is shown. Attention must be given to the change in kinematics and forces over time.

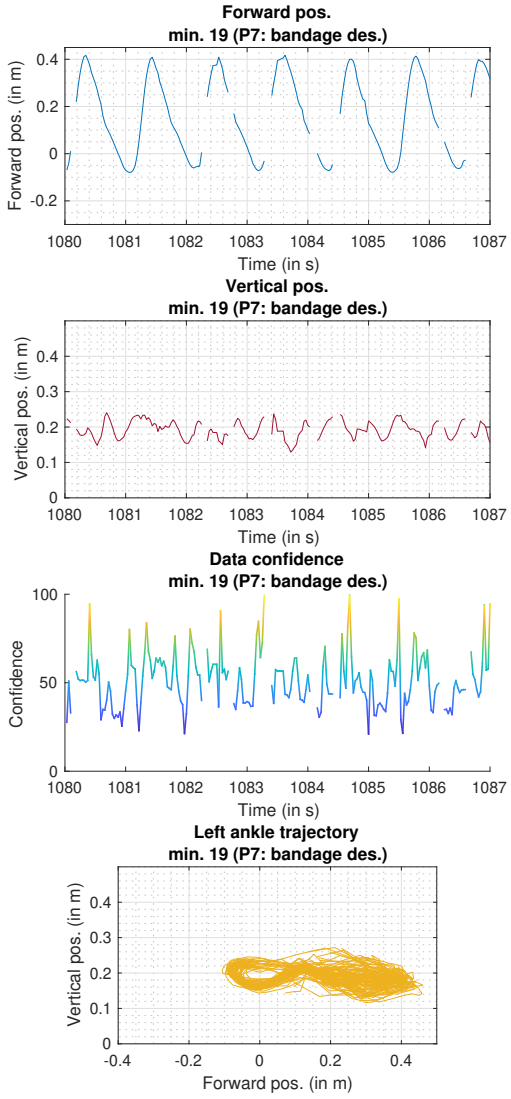


Fig. 28: Left Ankle kinematics. The confidence in the left ankle kinematics is lower than that in the right ankle. Additionally, there are more data gaps in the left ankle data. Therefore, the results in this paper are derived from right ankle data.

B. Questionnaire data

More information on the results of the questionnaires is provided in this sub-chapter. Figure 29 shows a bigger version of the box plots from the main text. The personalised icons displayed in each box plot and the differences in raking are more evident. Additionally, tables XII and XIII report the questionnaire scores of each participant.

TABLE XII: The table contains the scores for the User experience and system acceptance questionnaire per participant. U stands for usefulness, and S stands for satisfying scale. The scale range for user experience and system acceptance is -2 to +2.

Participant number	First tried attachment	Second tried attachment	User experience		MATE acceptance with Cuff		MATE acceptance with Bandage		Acceptance Cuff		Acceptance Bandage	
			Cuff	Bandage	U	S	U	S	U	S	U	S
1	Bandage	Cuff	1.52	0.52	1.2	2	1.8	1	2	2	1.2	0.75
2	Cuff	Bandage	0.8	0.6	0.8	1	1.4	1.75	1.6	2	-0.6	-0.25
3	Cuff	Bandage	0.55	-0.86	0.6	0.5	0.8	-0.5	1.4	-0.25	0.4	-0.5
4	Bandage	Cuff	0.9	0.67	0.8	1	0.6	0.75	1.2	0.75	1.4	1.25
5	Bandage	Cuff	0.81	0.9	1.2	1	1.4	0.5	0.8	0	1	0
6	Cuff	Bandage	0.29	0.43	0.2	0.25	0.6	0.25	0.6	0.5	0.4	0
7	Cuff	Bandage	0.1	0.76	0.6	0.25	1	1	0.6	0.5	1	1
8	Bandage	Cuff	1.1	-0.14	0.8	1	-0.8	-1	1.4	1.75	1	1.25

TABLE XIII: The table displays the scores for the Design Usability questionnaire for each participant. The scale ranges from 0 to 100. Additionally, the table lists the time it takes for participants to put on and take off both attachment designs from their legs, the donning and doffing time.

Participant number	First tried attachment	Second tried attachment	Participant Donning time (in s)		Participant Doffing time (in s)		Design usability	
			Cuff	Bandage	Cuff	Bandage	Cuff	Bandage
1	Bandage	Cuff	93	275	37	61	100	45
2	Cuff	Bandage	135	241	58	53	90	45
3	Cuff	Bandage	81	229	30	41	75	70
4	Bandage	Cuff	119	217	36	42	80	92,5
5	Bandage	Cuff	64	168	26	57	97,5	65
6	Cuff	Bandage	87	134	25	35	80	60
7	Cuff	Bandage	58	123	18	39	67,5	85
8	Bandage	Cuff	83	283	31	32	100	47,5

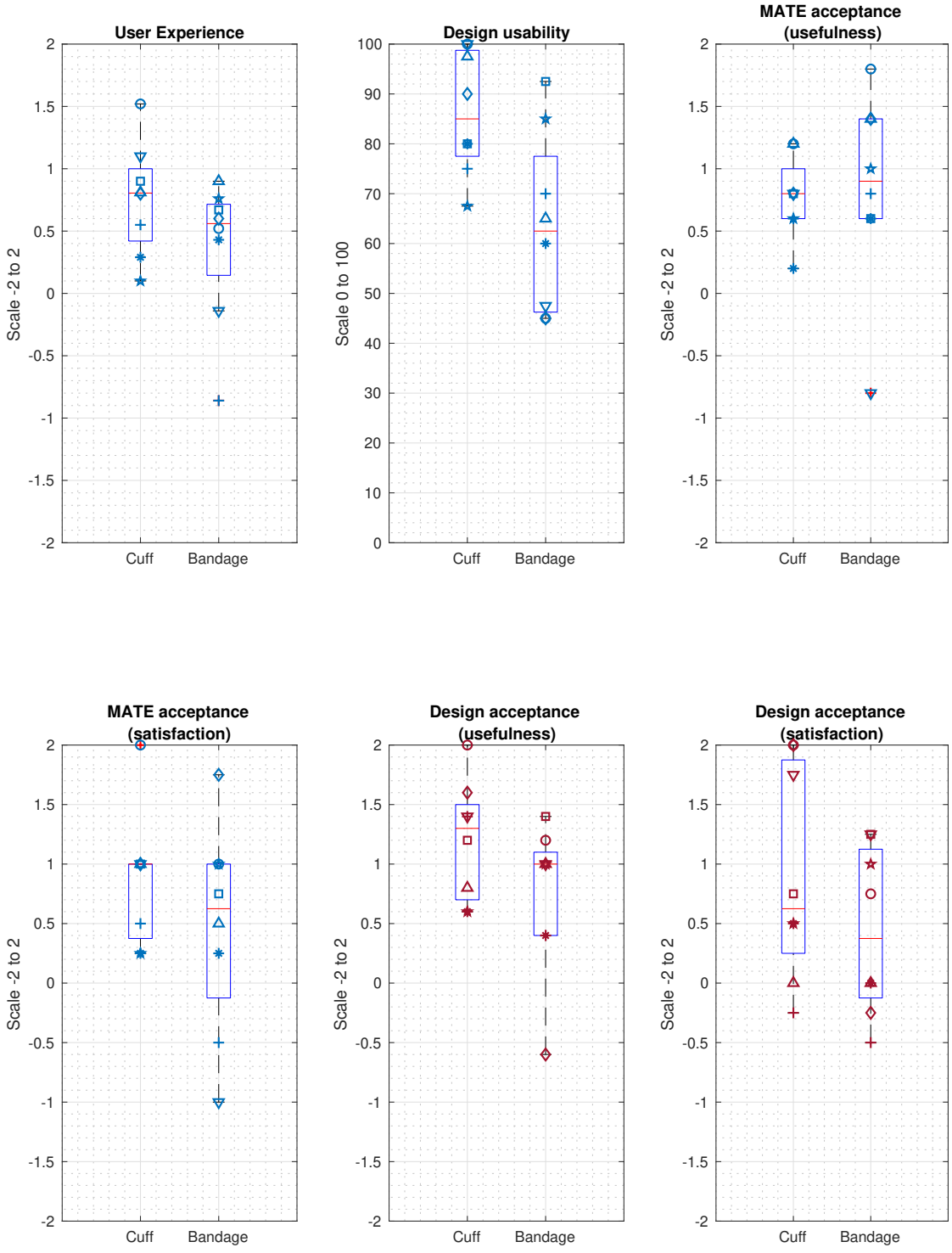


Fig. 29: This is an enlarged version of the box plots for the questionnaire results shown in Figure 7 in the main text. These box plots display the results of the User Experience, design usability, MATE acceptance, and design acceptance questionnaires. The median values are listed in table III in the main text.

C. Checklist data

During the experiment, in addition to the force data, kinematic data, and questionnaires, a checklist was completed that included demographic information of the participants, the order of putting on the attachment designs, the treadmill speed the participants walked at, and donning and doffing time of both models by the experimenter and the participants. In addition, design-related statements and comments were noted, and the height of the cuff design was measured before attaching the cables, after attaching the cables but before walking, and then walking with the cables still attached.

1) Demographic information, design order, treadmill speed, donning and doffing times, cuff height: All checklist data tables contain the participant numbers and the order of tried attachment designs.

- Demographic information of the participants; see table XIV.
- Treadmill speed per participant; see table XIV.
- Donning and doffing time by the experimenter; see table XV, and figures 30 and 31.
- Donning and doffing time by the participants; see table XIII, and figures 30 and 31.
- Cuff height differences before and after gait session; see table XVI.

TABLE XIV: In this table, checklist data of the participants' demographic information is displayed, together with the participant number, the order of the attachment designs, and the treadmill speed.

Participant number	Gender	Age (in years)	Height (in m)	Weight (in kg)	First tried attachment	Second tried attachment	Treadmill speed (in km/h)
1	Woman	63	1.60	79	Bandage	Cuff	3.0
2	Woman	23	1.65	70	Cuff	Bandage	3.0
3	Man	26	1.85	72	Cuff	Bandage	2.5
4	Man	61	1.73	79	Bandage	Cuff	3.5
5	Man	22	1.86	94	Bandage	Cuff	3.5
6	Man	24	1.91	86	Cuff	Bandage	3.5
7	Woman	24	1.67	70	Cuff	Bandage	3.0
8	Woman	60	1.63	71	Bandage	Cuff	2.5

TABLE XV: In this table, checklist data of the donning and doffing time by the experimenter is shown. The * at 18 seconds (participant No. 4) in the doffing time of bandage designs indicates that it is a doffing time without the protection pads of the design. The corresponding graphs are shown in figures 30 and 31.

Participant number	First tried attachment	Second tried attachment	Experimenter Donning time (in s)		Experimenter Doffing time (in s)	
			Cuff	Bandage	Cuff	Bandage
1	Bandage	Cuff	135	235	42	59
2	Cuff	Bandage	168	211	44	36
3	Cuff	Bandage	70	191	40	35
4	Bandage	Cuff	125	186	48	18*
5	Bandage	Cuff	86	157	28	44
6	Cuff	Bandage	76	134	33	41
7	Cuff	Bandage	61	127	37	41
8	Bandage	Cuff	74	127	30	36

TABLE XVI: The differences in cuff design height are reported in this table. Right after attaching the cuffs, the height between the treadmill/ground and the bottom of the right cuff is measured. This measurement is repeated right after attaching the cables but before the gait session, and finally right after the gait session with the cables still attached.

Participant number	Cuff height (in m)		
	After donning	Before 20 min walk, after cable attachment	After 20 min walk, cables still attached
1	-	0.15	-
2	0.11	0.12	0.13
3	0.13	0.14	0.18
4	0.17	0.18	0.19
5	0.13	0.14	0.175
6	0.15	0.15	0.175
7	0.11	0.12	0.125
8	0.13	0.135	0.155

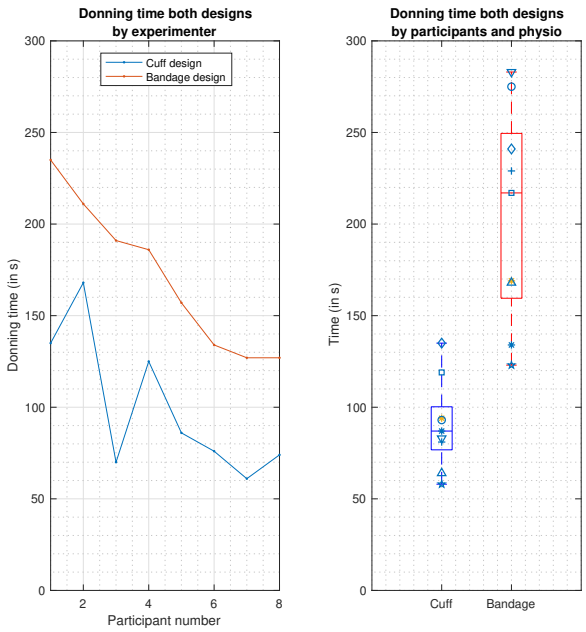


Fig. 30: Overview donning time of both designs. The left plot illustrates an overview of the experimenter's donning time (of both designs) with the cuff and bandage models. The right plot, the box plot, shows the donning time of the participants (blue figures) and the physiotherapist (yellow asterisks). The icons in the box plot are personalised per participant. The average donning time of the participants (+ physio) for the cuffs is 87 seconds, and 217 seconds for the bandage designs.

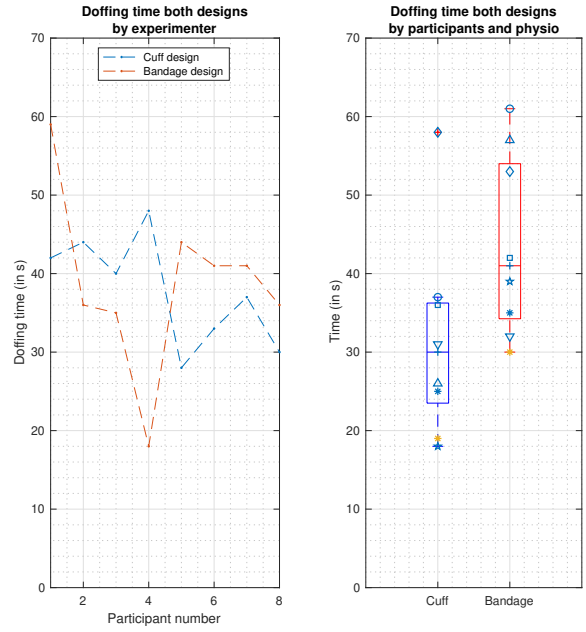


Fig. 31: Overview doffing time of both designs. The left plot illustrates an overview of the experimenter's doffing time (of both designs) with the cuff and bandage models. The steep drop to 18 seconds (participant No. 4) in the doffing time of bandage designs with the experimenter is a doffing time without the protection pads. The right plot, the box plot, shows the doffing time of the participants (blue figures) and the physiotherapist (yellow asterisks). The icons in the box plot are personalised per participant. The average doffing time of the participants (+ physio) for the cuffs is 30 seconds, and 41 seconds for the bandage designs.

2) Notes of participants: During the experiments, the participants' comments and statements were documented. Below, the comments are categorised by design. The quotes listed are originally communicated in Dutch. For this paper, they have been translated into English.

Notes and quotes on the Cuff design

- Participant 1 mentions she is walking her natural gait pattern but states she does not know whether enough support is being given.
- Participant 1 mentioned that using this model requires less focus on walking and that her whole body is more relaxed compared to walking with the bandage model.
- Heel strike and toe-off are hard for participant 2.
- Participants 2, 6, and 7 indicated that their gait pattern is now wider than normal. However, Participant 3 mentioned his gait pattern was less wide than normal.
- After 7 minutes, the walking became easier and more comfortable. (P2)
- Participants 3, 5, and 6 noticed the cuffs sliding upwards during the gait session, but no rotation was observed.
- Participant 3 can make his own heel strike, but toe-off is initiated earlier than normal. Also, Participant 3 took smaller steps than normal.
- Participant 4 is scuffing with his feet on the ground. He doesn't normally do that.
- Participant 4 indicated that because the cuffs are higher on the legs, it has less influence on his walking movement. The participant does have the idea that he is being guided by the MATE.
- Participant 6 mentioned that he slightly adjusted his gait pattern but felt comfortable with the changes. He was able to perform heel strikes and toe-offs. However, he took smaller steps to walk with the least resistance.
- After 10 minutes, participant 6 reported that his ankles were feeling heavier. He mentioned that the toe-off movement was different and suggested that his ankles accommodate this change or that it could be related to how the cuffs transfer forces to the body.
- Participant 7 mentioned that the cuffs are comfortable and feel like part of your body.
- Participant 7 indicates that her legs are pulled forward, and her ankles are guided. She can easily make her own toe-off, and at heel strike, she lands in the middle of her foot.
- Participant 7 walked with the design on her skin instead of wearing a sock in between. She mentioned that the fabric covering the cuff felt a bit sweaty, but it didn't chafe. She described it as feeling similar to a ski boot.
- Participant 8 can make her own heel strike and toe-off, and her legs are not pulled to the sides.
- Participant 8 states that her knee normally hurts. When walking with the cuff design, this pain disappeared completely, while with the bandage model, it got slightly worse.
- After 12.5 minutes, participant 8 indicates that walking starts to feel heavier.

- "With the bandage design, your foot is carried/lifted more; with this one, your toe can get stuck on the ground much more easily." (P1)
- "A combination would be ideal for me: something that is comfortable around the ankle but also has a strap under the heel to provide a lifting effect." (P1)
- "This design is much more comfortable, and walking is more natural than with the bandage design." (P1)
- "I feel like my legs are being corrected." (P3)
- "Walking becomes easier once you understand how the MATE works, and you are no longer working against pulling forces of the cables." (P3)
- "This feels like a regular walk, but the cuffs are more noticeable than the bandage design." (P4)
- "Easier to walk in your own gait pattern and make a heel strike and toe-off. Feet get lifted less high than with the bandage model" (P5)
- "This design (cuff) allows for more freedom of movement and reduces the widening of the legs. The other option (bandage) required less effort to walk." (P5)
- "The other design (bandage) provides more support. This cuff provides forward support, while the other also lifts your foot." (P5)
- "It's not a bother if you walk at a slower speed (1.5 km/h) with this design." (P5)
- "At a slower pace, walking is harder." (P6)
- "The MATE system helps with walking." (P6)
- Participant 7 after 10 sec: "it feels like I have to learn to walk again". After 1 min: "You get used to it fast. The first minute feels like you are being corrected (you feel like a puppet), but after that, I felt like I could control my own gait movements again."
- Participant 7 at 7.5 minutes: "Walking takes more effort, mainly because you walk a bit wider than normal." She indicated that she is trying to counteract the outward pull.

- "This walks much better than the other one. Now I can just walk like I normally do." (P8)
- "My legs are now guided, and there is no feeling of 'fighting against the cables."' (P8)
- "Now, it is very nice that your feet feel like your feet again."
- "Because the bandage design looks nice and professional, it immediately gets your preference, while ultimately, the cuff is much easier to put on. The bandage design does not look like an aid, and the cuff does." (P8)

Notes and quotes on the Bandage design

- Participant 1 can make her own gait pattern.
- Heel strike and toe-off are hard for participants 1, 3, 4, and 8.
- The protection pad with one sealing strap is most comfortable around the ankle, according to Participant 3.
- Participant 4 wore leather boots, so protection around the ankles during the bandage design was considered unnecessary. However, the participant mentioned adding the pads for more comfort at the end of the gait session.
- Participants 4, 5, and 8 indicated that their gait patterns are now wider than normal.
- Participant 4 makes smaller steps than normal.
- Participant 4 does not get the impression that he is being supported by the MATE.
- The buckle on the back of the designs rested on the Achilles tendon instead of the shoe while walking, causing slight irritation. (P6)
- Participant 6 indicated that he is more in tune with the design but struggles to make heel strikes. Instead, he is landing more in the middle of my foot.
- "This design is more pleasant. Even though you tiptoe more, this feels more solid." Participant 6 said he dares to walk with his eyes closed, which was less so with the cuff design.
- Walking at a slower speed (1.5 km/h) was not necessarily heavier. (P6, P7)
- Participant 7 mentions she walks less wide-legged than with the cuff.
- Participant 7 feels safer with this model because she has more control. With the cuff, she felt she could trip if she wasn't paying attention and had to focus. With the bandage model, "walking takes more effort, but you have more control and freedom."
- Later in the gait session, heel strike and toe-off can be made by Participant 7.
- Participant 8 mentioned that walking with the bandage model became easier with time. After 8.5 minutes, she was able to make her heel strike and toe-off again. By 10 minutes, she mentioned that she could easily maintain this. At 12 minutes, she said she could complete her normal gait cycle again. By 13.5 minutes, she mentioned feeling more guided than controlled and suggested that practising with the MATE would lead to better results.
- Participant 8 says she walks very differently from her normal gait pattern, indicating she is pulled forward and sideways by the cables.
- Participant 8 likes the bandage design better and more professional without pads and says, "You feel a bit of a dork with the pads, and without it, you feel like an athlete."
- Participant 8 mentioned that she now expends less energy while walking compared to the beginning of the gait session, although it's not at all considered energy-saving.

- "My legs are being lifted." (P1, P3, P5, P6)
- "When you're in a flow, walking is easy. However, you start walking like a soldier when you lose concentration." (P1)
- "It takes some getting used to at the beginning. After about 3 steps, you understand how it works, and after 3 minutes, you can walk normally." (P1)
- "This is less comfortable. You didn't really feel the other one, and with this one, you feel pressure on your ankles. It is nicer that your foot is lifted with this model. Now there are no problems with heel strike and toe-off." (P2)
- "What a hassle, when you think you're done putting it on, those protection pads still have to be added..." (P2)
- "This one is less relaxed than the cuff design because you cannot make your own gait pattern." (P3)
- "The abrupt pulling force at toe-off slightly puts you off balance. And this increases as you increase your step length." (P4)
- "It takes more energy to walk than normal. You have to pay more attention to your balance." (P4)
- "I feel like I'm walking a little differently, but I don't have to put as much effort into lifting my foot." (P5)
- "It feels like you are being pulled forward." (P5)
- "It is easier to walk at 3.5 km/h than at 1.5 km/h because you want to move your leg forward faster than possible at 1.5 km/h." (P5)
- "It feels like I walk more clumsily and stomp more." (P6)
- "Your feet are controlled, and you have to apply more force to go down with your foot." (P6)
- "This design does really define your gait cycle. Making your own gait cycle takes effort, and following the MATE takes no effort and is even energy-saving." (P6)

- "Odd, but more freedom of movement. Feet are being guided" (P7)
- "When I take a step, I land on my heel more than with the cuff design. The feet are pulled up more, so I need more force from my upper legs to put them on the ground." (P7)
- "Walking requires more effort than with the cuff." (P7)
- "Design is comfortable. It's nice that you don't have warm fabric around your leg." (P7)
- Participant 8 referring to gait pattern: "I feel like a duck."
- "Feet are controlled" and "the cables have control over the legs." (P8)
- "It costs energy to make your gait cycle." (P8)
- "The protective pads are palpable on the right foot, but not on the left." (P8)
- "As an elder, you are fighting with the MATE." (P8)

APPENDIX E
DESIGN ASSESSMENT

The Unit Infection Prevention (UNIP) specialisation at Erasmus MC focuses on infection control and prevention. It provides the right preventive measures to prevent the spread of all kinds of microorganisms and keep healthcare-associated infections low. One of the measures is proper device cleaning [38]. Create4Care helps solve practice problems through smart technical innovations [39].

A. Interview UNIP

At the start of the interview, the designs and their donning and doffing were explained and demonstrated. Then, the experiment's hygiene measures were explained, and a free discussion followed.

Hygiene rules for designs like these:
- Every time a physiotherapist does the donning or doffing of the designs, he/she will wash their hands thoroughly. This limits the chance of transferring bacteria from one patient to another.
- When designs are washed by hand, it is done in the following methods.
    - o When there is blood on the designs: cleaning wipes with a 70% alcohol solution or cleaning wipes with 250 ppm chlorine solution.
    - o When there is no blood, use a damp microfiber cloth.
- There is a chart to check what is the best way to clean a product in specific situations.
- Elements of a design can be cleaned separately.

Preferred materials
- Plastic materials are most often used, but whether this material is an option is design-dependent.
- Rubbers could be used if it is not porous and tolerant to alcohol.
- The material used should be resistant to alcohol and chlorine.

Common design mistakes
- Using rough textured material. This material is hard to clean and can easily transfer bacteria between users.
- Using materials that cannot withstand chlorine or alcohol. For instance, rubbers often become hard, and often, aluminium cannot withstand alcohol either.
- Even with large surfaces, there should be no gaps, wrinkles, or open ends (even if it is not even close to where the human touches the design).

Sustainability
- The preferred option is the most sustainable one, where a design can be used by multiple patients.

1) Recommendations for the next iterations.:

Both designs
- During the experiment, the participants wore their own high socks to prevent shear spots and for hygiene. However, the interview reveals that this does not help increase hygiene as bacteria go straight through them. Plus, bacteria are also on clothes and not just skin.
- If the design is personalised, it is essentially like garments. There are slightly less strict hygiene rules than when a design is shared between patients, but it must still be properly cleanable, as patients might otherwise re-infect themselves with bacteria. This is also why foam should not be allowed, even if a design is specific to the user.
- Now, the current designs can be used one time only and then need to be thrown away due to the lack of washability.

Bandage design

Materials:
- Synthetic leather is in itself a good choice. However, the seams should be sealed so that no gaps are left in the design where bacteria can get between. In addition, the material used should be resistant to alcohol and chlorine.
- Velcro is a no-go. It's only acceptable if you can wash the design at 60 degrees Celsius.
- The straps should made of more plastic materials, as they are too porous now.
- Foam is a definite no-go; it is a huge bacterial hotbed and cannot be cleaned properly (as is it not washable). However, foam can be used if it is on the inside of a fully sealed part of the design. As an alternative, hard but still flexible plastic could be used. But again, this material should not be porous.

- From the interview, it became evident that some already existing plastic bands might be a good replacement for the currently used straps.
- Materials should be hard, smooth surface/finish, non-porous.

Model:

- The protection pad with the two fasteners AND the protection pad with the slit from the bottom up to the middle of the design have the most potential for the sake of hygiene. However, this only applies under the condition that they are fully sealed. If a part is completely sealed and fully waterproof, foam is allowed inside.
- The model should be easily wipeable/washable and without ridges and wrinkles.

Cuff design

Materials:

- Foam is a NO unless completely sealed off.
- Can the synthetic leather withstand alcohol/chlorine? Also, the inside of the used synthetic leather is not hygienic.
- Also see information under 'bandage design'.

Model:

- The ridges in the cover of the front component of the cuff are an immediate no… dirt gets stuck in there. Generally, covers/sleeves for a design are okay, but then foam should not be used for the design.
- Velcro on aluminium cannot be machine-washed and is therefore not hygienic.

2) Extension on the hygiene requirements and wishes:

Extension of the list of requirements for the next iterations of the designs focused on hygiene.

- Materials should be hard, smooth surface/finish, non-porous.
- The model should be easily wipeable/washable and without ridges and wrinkles.
- The materials should be resistant to alcohol and chlorine or washable at 60 degrees Celsius.
- Reduce the number of separate parts in a design. The more individual parts, the higher the chances of losing one of them.
- Check whether patients are allergic to the materials used.

Extension of the wish list, focused on hygiene

- Easily re-usable designs between patients.
- The design operation must be easy to understand.
- There must be good instructions manual on how to clean the design. Should the design be taken apart for the cleaning process, HOW should it be taken apart, and in WHICH order? The same goes for putting it back together again.

3) Next steps: The Create 4 Care department at Erasmus MC brainstorms and designs medical products. For future iterations of the designs, one could contact them for feedback and inspiration.

B. Interview Create4Care

After the interview with UNIP, I was introduced to some of the people in the Erasmus MC department, 'Create 4 Care'. I showed them the prototypes, and we discussed possible improvements on both designs. Not everything is written down, as there was a lot of overlap with the UNIP interview.

Both designs

- Velcro is a no-go, but silicone cable ties/tie-wraps might be a good alternative. Other possible options to replace Velcro:
  - o Magnets
  - o Belt buckles
- Foam attracts dirt and it easily soaks in dirt. However, using it when it is completely sealed off might be possible.
- Medical synthetic leather is a material that is resistant to the proper cleaning products. The material is already being used in o.a. hospitals.
- Once a design is taken home by the patient, hygiene cannot be controlled as patients cannot be controlled. However, a wider range of materials can be chosen if only one patient uses the design.

Bandage design

- Do not use:
  - o Velcro
  - o Porous fabric

- Designs cannot always be washed on location. It might get lost (or parts of the design), or locations don't have a washing machine.
- The Create 4 Care team remembers working on a project in the past focused on custom-made designs. However, they ran into the same design problems (as with the bandage design) because of the Velcro. The team says there are clip systems that can be customised; think claps on rucksacks. However, those claps also contain corners and nooks that can get dirt in them.

Cuff design

- It is preferable for the design to consist of one or two parts, as parts often get lost. An option would be for the cover to stay around the cuff itself but be cleaned properly as a whole.
- The team mentions: Suppose the aluminium band could unroll completely, eliminating wrinkles and dimples and allowing it to be cleaned properly. Or just a separate foam cushion, which is securely sealed, that can be tucked into something C-shaped.
- As mentioned earlier, using foam is not ideal, so there should be a good replacement for this material. Additionally, Nylon could be used instead of aluminium (as aluminium often does not tolerate cleaning products well). However, Nylon is more fragile.

APPENDIX F
EXPERIMENTAL SET-UP

A. MATE

This study employs the framework, elastic cables, and pulley construction from the previous study on the MATE. In figure 32, a Matlab visualisation of the MATE is shown in which the position of the four different cables is displayed, and in figure 33, the measurements for the pulley configuration are given. For further information on the frame, cable, and pulley construction, please refer to Kieft's paper 32.

- Frames (different lengths of aluminium extruded profiles from Item, profile 8 40x40 artikel-nr. 7.0.000.09 [11])
- Elastic cables (GAMMA, 8mm, product code 339015 [11])
- Pulleys (GAMMA product code 8712513061851 [11])

In Kieft's paper, the overall minimal cable lengths were presented in that paper's Table 2. However, when those cable lengths were employed in the MATE construction rebuild for the experiment in this study, the cables were slack. When the cables were found at the beginning of this project, the cables had an unstretched length of:

- Front high cable = 1.02 meter.
- Front low cable = 1.62 meter.
- Back high cable = 1.30 meter.
- Back low cable = 2.02 meter.

These cable lengths did not have any slack during use, so they were used as the standard lengths. However, during the experiment, each cable was equipped with three carabiners, a force sensor, and a cable tension hook between the cable and its attachment point. These items collectively added 33 cm to the length of each cable, so this additional length had to be subtracted from the original cable measurements. As a result, the standard unstretched cable lengths used during the experiments were:

- Front high cable = 0.69 meter.
- Front low cable = 1.29 meter.
- Back high cable = 0.97 meter.
- Back low cable = 1.69 meter.

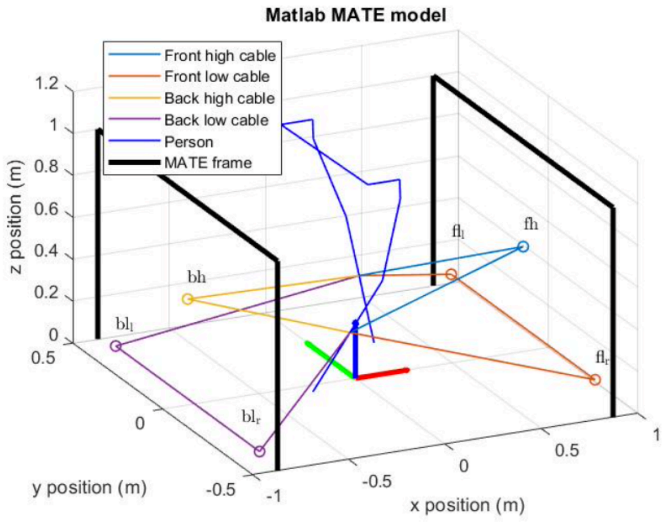


Fig. 32: This is figure 2 in the paper of Kieft [11], showing a Matlab visualisation of the MATE and the cable construction.

However, before a participant started walking with the MATE for the first time in the experiment, the experimenter asked if the pulling force on the legs was comfortable and not pulling so hard that it would affect the participant's natural gait pattern during walking. If the pulling force was perceived as too tight by the participant, the cables were elongated until the pulling force was perceived as comfortable. The participant used the same cable lengths in both the gait sessions, during which he/she walked with the MATE. In Table X, the cable lengths per participant are displayed together with the pulling force anterior and posterior on the attachment designs during a standstill in the middle of the treadmill.

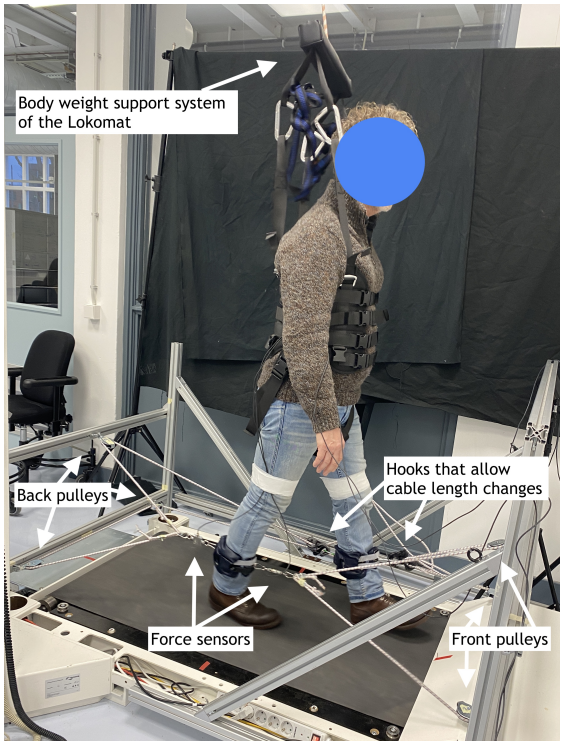
PULLEY CONFIGURATION, OPTIMAL FROM [20] AND THE IMPLEMENTED VALUES IN THE BUILT MATE

parameters	x_{fh}	z_{fh}	x_{fl_l}	z_{fl_l}	x_{fl_r}	z_{fl_r}	x_{bh}	z_{bh}	x_{bl_l}	z_{bl_l}	x_{bl_r}	z_{bl_r}
optimal	0.00 m	0.50 m	0.40 m	0.10 m	−0.40 m	0.10 m	0.00 m	0.80 m	−0.40 m	0.10 m	0.40 m	0.10 m
implemented	0.00 m	0.50 m	0.40 m	0.13 m	−0.40 m	0.13 m	0.00 m	0.50 m	−0.40 m	0.10 m	0.40 m	0.10 m

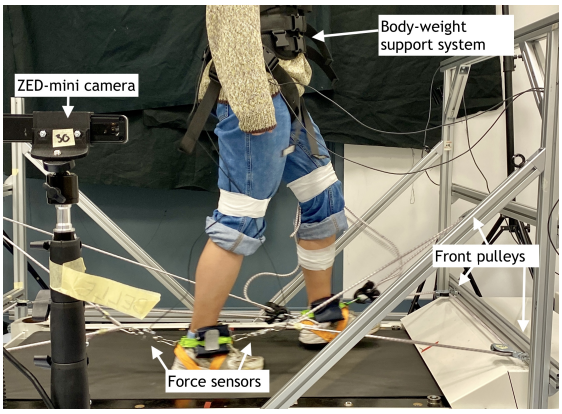
Fig. 33: This is table 1 in the paper of Kieft [11], showing the measurements of the pulley configuration. The implemented values are highlighted because those measurements were also taken for the study in this paper.

B. Attachment Designs & test set-up

The attachment designs, the cuff and bandage, are explained in detail in Appendix A-D, as is the connection between the attachments to the cables. In figure 34, two overviews of the test set-up of the experiment in this study are shown, in which the position of the pulleys, force sensors, ZED-mini camera, cable elongation/tension hooks, and body weight support system of the Lokomat are displayed.



(a) An overview of the test set-up is shown in this figure. The cables, the pulleys, the body-weight support system, the force sensors and cable elongation hooks are displayed.



(b) An additional overview of the test set-up is shown in this figure, where the position of the ZED-mini is displayed. The position of the ZED-mini to the treadmill is noted in Table XVII.

Fig. 34: Overview experimental test set-up.

C. Force sensors

1) Where are the sensors attached?: The load sensors are connected between the ends of the cables on the right and the right attachment. One load sensor is positioned between the back cables and the back of the right attachment, while the other load sensor is positioned between the front cables and the front of the right attachment.

2) How are the forces recorded?: The two load sensors, KDC 40S 1000N, have a range of ±1000N per sensor. The load sensors are both connected to an analogue amplifier LCV-U10 10V, which amplifies the sensor data to a range of ±10V. (The two amplifiers are powered by a 12V adapter.) This voltage signal is measured by an Arduino that sends the corresponding force value via serial communication to the computer, connected via a USB cable, where the data is stored in a .csv file. In the force calculations of Appendix B, it was calculated that during normal gait, there could be around 150N on each force sensor during the experiment. Knowing that the force sensors can sustain up to 1000N, the 150 N will not cause a problem. Also, the eyebolts (GAMMA Product code 432621) used on the sides of the load sensor to connect them to the cables can withstand up to 200kg, which is enough for the force used in this experiment.

3) Force sensor calibration: The force sensors were calibrated with a 20 kg weight from the IWS workshop before the first experiment and after the final experiment to verify that 20.1 kg corresponded to 197.2 N. Before the trials, the 20.1 kg weight (20kg weight + 100 grams hook) resulted in 196.9 N at the front sensor and 190.6 N at the back sensor. Subsequently, after the experiments, the same 20.1 kg weight was measured at 178.1 N for the front sensor and 199.4 N for the back sensor. Some fluctuations are observed in the responses; however, both force sensors, either before the experiments or after, give a response that is close to the desired 197.2 N. For a subsequent study, multiple trials can be conducted during calibration to test the force sensors' accuracy better. In the study in this paper, the force sensors were presumed to be accurate, and no adjustments were made to the data analysis based on this force calibration.

Additionally, a second force sensor calibration was performed. At the beginning and end of each participant's experiment, a zero point was determined with the force sensors resting on the table without any external force. Since these zero points did not consistently indicate 0 N, the offset per participant was identified and incorporated into that participant's force data.

D. ZED mini camera

1) Motion capture: During the gait sessions of the experiment, kinematic data is recorded using the ZED mini camera from StereoLabs [40]. This depth camera is placed in the sagittal plane on the right side of the body. This way, there is a clear view of the participant using the MATE and the attachment designs on the right leg, so the participant's kinematic data can

be obtained. No markers are placed on the participant's body as the ZED mini depth camera has built-in motion trackers; see figure 35. Body format BODY18 is used in this study, where only markers 1 (neck), 8 (right hip), 9 (right knee), 10 (right ankle), 11 (left hip), 12 (left knee), and 13 (left ankle) are used [41]. From these key points, the x, y, and z positions and the detection confidence are obtained for analyses in Matlab. Camera calibration was already completed in-house [42].

2) How are the force sensors and the motion capture system synced?: Both the force sensors and kinematic data are synchronised with the time and number of frames. Additionally, real-time force readings are displayed in the top left corner of the ZED-mini video. The Python script used for processing the data during the experiments can be found in Appendix H.

3) Distance between the ZED-mini camera and the frame: The position of the ZED mini camera according to the frame was adjusted according to the length and size of the participant. It was placed in a total of three different positions during the experiment. The camera was positioned so that the participant's body (not the head) was in full sight at all times (together with the pulleys as far as was made possible by the frames of the MATE and the Lokomat) and so that the ZED-mini marker on the participant's neck was still visible, as this ensured that the markers could follow the participant's body more precisely while walking. However, this last statement was an assumption based on testing the set-up and the ZED-mini camera and was not investigated in more detail. The position of the ZED mini camera for the different participants is shown in Table XVII. The ZED-mini camera was placed 88 cm above the ground for all participants.

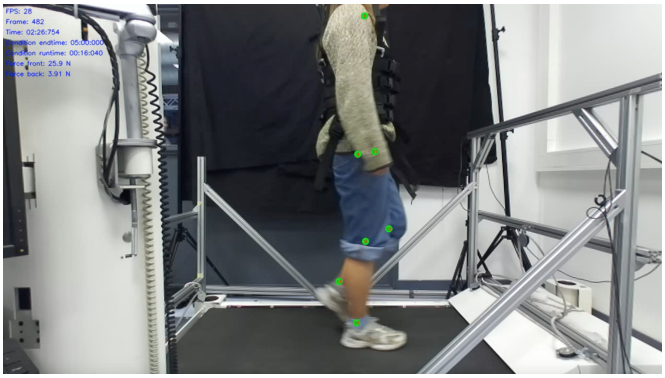


Fig. 35: Snapshot of the ZED-mini's image with body markers during baseline.

TABLE XVII: This table shows the distance from the ZED-mini camera to the frame on the participant's right side while walking.

Distance from ZED-mini camera to frame									
Participant	1	2	3	4	5	6	7	8	Case study Two: physiotherapist
Distance (in m)	1,03	1,03	1,36	1,10	1,36	1,36	1,03	1,03	1,36

E. Questionnaire forms

Three different questionnaires were used to obtain the participants' user experience during the gait sessions, their acceptance of the MATE with the designs, their acceptance of the designs themselves, and finally, information about the design usability.

1) User Experience Questionnaire: The user experience of interactive products is evaluated using the User Experience Questionnaire [20]. In this study, we modified the existing user experience questionnaire to elicit more targeted feedback on the user experience of the MATE. After the 20-minute gait session with the MATE and each design, the customised User Experience questionnaire was completed. The customised questionnaire can be seen in figures 36 and 37.

2) Acceptance Scale questionnaire: Right after the User Experience questionnaire, the Van der Laan Acceptance questionnaire on the MATE with the design was completed, followed by the Van der Laan Acceptance questionnaire on the design itself; see figure 38. The system acceptance scale assesses the system (in this study, either the MATE with the attachment design or the attachment design itself) on two dimensions: a Usefulness scale and a Satisfying scale [21].

3) System Usability Scale (SUS) questionnaire: The System Usability Scale is a reliable tool for assessing the usability of systems on a global scale [22][43]. Sauro states that the average SUS score is 68 [44]. In this experiment, the SUS questionnaire was completed after the participant did the donning and doffing of the designs. For the SUS questionnaire, see figure 39

F. Checklist

We collected data using force sensors, participant kinematics, videos, and questionnaires during the gait sessions. In addition, we applied the 'thinking-aloud' protocol, where the experimenter engaged in conversation with the participant and the participant was encouraged to verbalise their thoughts [45]. We also had the experimenter and participants don and doff the designs to supplement our user observations.

Throughout the experiment, a checklist was maintained for each participant to ensure the accurate execution of tasks. This document also recorded the participant's demographic information, the design order, comments and observations during the gait sessions with the MATE, cable lengths, treadmill speed, place of the ZED-mini camera, and the time taken for both the participant and the experimenter to don and doff the attachment designs. The checklist is shown in figures 40, 41, and 42.

Questionnaire: ..... design

Participant number: .....

User Experience

Receiving support from the MATE system when stepping forward, was:
Comfortable |__|__|__|__|__| Uncomfortable

The MATE was influencing my gait cycle (e.g., heel strike, toe-off, mid-swing, midstance) in a good/bad way. (Helping you walk.)
Bad |__|__|__|__|__| Good

The attachment designs felt securely connected to my legs.
Secure |__|__|__|__|__| Not secure

Did your normal way of walking get affected by the MATE system?
Affected |__|__|__|__|__| Unaffected

I can imagine that walking with the MATE system during gait training is...:
Undesirable |__|__|__|__|__| Desirable

I was engaged while walking with the MATE system:
Raising Alertness |__|__|__|__|__| Sleep-inducing

The MATE system is useful/useless for saving energy during the 20-minute walk:
Useful |__|__|__|__|__| Useless

Did the MATE system help you walk more effortlessly?
The MATE was:
Assisting |__|__|__|__|__| Resisting

During the 20-minute walk, did you like the supporting forces of the MATE?
Irritating |__|__|__|__|__| Likeable

Were you able to complete your own gait cycle comfortably (e.g., heel strike, toe-off, midstance, mid-swing)? In other words: the MATE helping me complete my gait cycle was:
Effective |__|__|__|__|__| Superfluous

The MATE system was causing friction between my body and the attachment?
Friction |__|__|__|__|__| Frictionless

By the end of the 20-minute walk, the MATE was:
Noticeable |__|__|__|__|__| Unnoticeable

Walking with the MATE system was:
Enjoyable |__|__|__|__|__| Annoying

Fig. 36: Questionnaires User experience page One

During the 20-minute walk, the attachments connected to your body felt:
Pleasant |__|__|__|__|__| Unpleasant

The support of the MATE system during walking was:
Nice |__|__|__|__|__| Annoying

I adapted to/fought against the support the MATE system exerted in my legs.
Adapted to|__|__|__|__|__| Fought against

The attachment design is:
Practical |__|__|__|__|__| Impractical

The received support of the MATE system was:
Efficient |__|__|__|__|__| Inefficient

The attachment designs were pain-free/painful to wear?
Pain free |__|__|__|__|__| Painful

Receiving support from the MATE to lift my feet during walking was:
Practical |__|__|__|__|__| Impractical

The MATE system makes it easier to walk.
Easier |__|__|__|__|__| Harder

Fig. 37: Questionnaires User experience page Two

System acceptance

I find the MATE system ..... (please tick a box on every line).

1 Useful|__|__|__|__|__|__|Useless
2 Pleasant|__|__|__|__|__|__|Unpleasant
3 Bad|__|__|__|__|__|__|Good
4 Nice|__|__|__|__|__|__|Annoying
5 Effective|__|__|__|__|__|__|Superfluous
6 Irritating|__|__|__|__|__|__|Likeable
7 Assisting|__|__|__|__|__|__|Worthless
8 Undesirable|__|__|__|__|__|__|Desirable
9 Raising Alertness|__|__|__|__|__|__|Sleep-inducing

I find the attachment design ..... (please tick a box on every line).

1 Useful|__|__|__|__|__|__|Useless
2 Pleasant|__|__|__|__|__|__|Unpleasant
3 Bad|__|__|__|__|__|__|Good
4 Nice|__|__|__|__|__|__|Annoying
5 Effective|__|__|__|__|__|__|Superfluous
6 Irritating|__|__|__|__|__|__|Likeable
7 Assisting|__|__|__|__|__|__|Worthless
8 Undesirable|__|__|__|__|__|__|Desirable
9 Raising Alertness|__|__|__|__|__|__|Sleep-inducing

Fig. 38: Questionnaires System acceptance

Usability of the attachment designs

The system: cuff design / bandage design

1. I would like to use this system frequently.
Strongly disagree O O O O O strongly agree

2. The system was unnecessarily complex.
Strongly disagree O O O O O strongly agree

3. The system was easy to use.
Strongly disagree O O O O O strongly agree

4. I would need the support of a technical person to be able to use this system.
Strongly disagree O O O O O strongly agree

5. The various functions in this system were well integrated.
Strongly disagree O O O O O strongly agree

6. There was too much inconsistency in this system.
Strongly disagree O O O O O strongly agree

7. Most people would learn to use this system very quickly.
Strongly disagree O O O O O strongly agree

8. The system is very cumbersome to use.
Strongly disagree O O O O O strongly agree

9. I felt very confident using the system.
Strongly disagree O O O O O strongly agree

10. I needed to learn a lot of things before I could get going with this system.
Strongly disagree O O O O O strongly agree

Fig. 39: Questionnaires Design usability

Checklist/script experiment (detailed protocol)

PER PARTICIPANT

Notes:

- The participant should have a dominant right leg, for consistency in the experiment.
- The attachment designs are alternately used first or second. (Check table with randomized order.)
- The participant is allowed to bring headphones to the experiment to deal with the construction noise. If so, they are only allowed to listen to white noise and not music. However, it is preferred that the participant focuses on the experiment and talks to the experimenter about their experience.
- There is no mechanical safety stop in this design because the MATE does not exert forces so high on the participant that a safety stop is necessary.
- Ask participants to wear high socks, so the attachment designs will not touch the skin of the participants. Socks will be provided if necessary.
- Calibrate the force sensors before the first participant (with a known weight of 20kg). After the last participant recreate the same test to check if the result is still the same.
- Have an extra person in the experiment room at all times.
- Participant should go to the bathroom before the start of the experiment.

Before participant arrives:

- ☐ Check the body weight support system (BWS) for damage.
- ☐ Check BWS harness on damage.
- ☐ Install frame MATE.
- ☐ Check if all the pulleys are in the correct spot.
- ☐ Check both designs for damage.
- ☐ Check elastic cables for damage.
- ☐ Install elastic cables in the frame.
- ☐ Make or check marked points on the sides of the treadmill that indicate the middle of the treadmill.
- ☐ Open all needed documents and programs on the computer.
- ☐ Check the force sensors. (hardware-related)
- ☐ Calibrate the force sensors at the beginning of each day (done per participant during test)
- ☐ Check the video.
  Length treadmill to the front pole of the camera stand:.....
- ☐ Check if the video recording is focused on the right frame. In the sagittal plane on the left side of the participant, are the following things still visible when the feet are ± 60 cm from the start place: Hip, knees, ankles, attachment designs. Mark the spot where the camera will stand during the experiments.
- ☐ Make sure adapters are nearby/put in both the used laptops.
- ☐ Make sure **both questionnaires** are present (multiple copies).
- ☐ Make sure **informed consent** papers are present (multiple copies).
- ☐ Make sure this **checklist** is present (multiple copies).
- ☐ Make sure the participant number (1-8) is noted on the documents and in the script.
- ☐ Make sure there are two extra pens.
- ☐ Make sure the set-up is clean and clear.
- ☐ Optional: notes: .....

(a) Page one.

Participant arrives:

- ☐ **Participant number**: .....
- ☐ The experiment set-up is explained to the participant.
- ☐ Accentuate the location of the stop button.
- ☐ The experiment protocol is explained to the participant.
- ☐ The participant reads and signs the informed consent form, on paper.
- ☐ Note the following **demographic information of the participant**.

  Gender:.....
  Age (in years):.....
  Length (in m):.....
  Weight (in kg): .....
  Lengths elastic cables:
  - Front high cable:.....
  - Front low cable:.....
  - Back high cable:.....
  - Back low cable:.....
- ☐ The donning and doffing of the two designs are explained and demonstrated to the participant.
- ☐ Check if the participant understands.
- ☐ Optional: notes: .....

Part one of the experiment starts, where the participant will walk on the treadmill without the MATE for 5 minutes.

- ☐ The BWS harness is put on the participant. And lock the participant in the framework of the BWS.
- ☐ Adapt BWS to the participant.
- ☐ Make sure elastic cables are secured to the MATE frame, and not troubling the participant.
- ☐ Participant will stand in the middle of the treadmill, indicated by the marked points.
- ☐ Check if the video recording is focused on the right frame. In the sagittal plane on the left side of the participant, are the following things still visible when the feet are ± 60 cm from the start place: Hip, knees, ankles, attachment designs.
- ☐ Ask if the participant is comfortable.
- ☐ Optional: notes: .....
- ☐ Start the treadmill.
- ☐ Obtain the preferred walking speed of the participant.
  Preferred walking speed:.....
- ☐ Start video + force sensors (there is no force on the force sensors in this part, so the mean of this data will be used as the null measurement).
  The participant now walks on a treadmill.
- ☐ Tell the participant to look forward as much as possible.
- ☐ The participant walks on the treadmill in the preferred walking speed for 5 minutes.

(b) Page two.

- ☐ Optional: notes. .....
- ☐ After 5 minutes, stop the treadmill and stop the video.
- ☐ Unlock the participant from the BWS frame (the participant keeps on the harness).

- ☐ The participant will sit down on a chair and can take a rest if needed (minimum rest time is 5 minutes).
- ☐ Check if all files are saved in the designated folders. (.csv, and .mp4)
- ☐ After the moment of rest, the experimenter puts on the first attachment design on the participant, on both ankles. The donning time will be recorded with a stopwatch by the experimenter and noted.
  Attachment design round 2:.....
  Donning time experimenter: .....
- ☐ Optional: notes. .....
- ☐ Before entering the treadmill with the first design around both ankles, they are checked by the experimenter, for a safety check.

(c) Page three.

Part two of the experiment starts, where the participant walks on the treadmill for 20 minutes with the first design.

When cuff, measure the length from ground to bottom line of the cuff before and after walking for 20 min. on the treadmill.

CUFF HEIGHT BEFORE 20 MIN WALK: :.....
CUFF HEIGHT AFTER 20 MIN WALK:.....

- ☐ Lock the participant back in the BWS.
- ☐ The participant will stand still in the middle (indicated by the marked points) of the treadmill and MATE will be attached to the design on both ankles.
- ☐ Check if the force sensors are attached to the left leg, and the tension hooks are on the right leg.
- ☐ Change the length of the cables until the preferred pretension is reached.
- ☐ Check if the video recording is focused on the right frame. In the sagittal plane on the left side of the participant, are the following things still visible when the feet are ± 60 cm from the start place: Hip, knees, ankles, attachment designs.
- ☐ Check if the cables of the force sensor, and the ends of the elastic cables, are fasted to the participant's legs with Coban tape. Without taking away the working quality of the cables or influencing the participant's gait cycle.
- ☐ Ask if the participant is troubled by the cables. If yes, repeat the checkpoint above.
- ☐ Optional: notes: .....
- ☐ Ask if the participant is ready to start.
- ☐ Start measuring the force sensors, while the participant is still at a standstill at the (middle) place indicated by the markers.
- ☐ Measure force sensors for 10 seconds (to figure out the offset later).
- ☐ Stop measuring the force sensors.
- ☐ Ask if the participant is ready to start walking.
- ☐ Start the treadmill.
- ☐ Set walking speed to the preferred walking speed measured in part 1.
- ☐ Start the video and force sensors.
- ☐ Check if the video and force sensors are being recorded.
  The participant now walks on a treadmill.
- ☐ Tell the participant to look forward as much as possible.
- ☐ During the experiment keep checking if the pulleys are in the right spot and if the participant walks in the middle of the treadmill (indicated by the marked points).
- ☐ Optional: notes: .....
- ☐ Stop video and force sensor recording at minute 20 of the experiment (timing starts once the participant is walking in the preferred speed).
- ☐ Stop the treadmill right after.
- ☐ The participant will stand still in de middle of the treadmill (indicated by the markers).

(d) Page four.

Fig. 40: The experiment checklist completed for every participant.

- ☐ Start measuring the force sensors (while the participant is still at a standstill at the (middle) place indicated by the markers).
- ☐ Measure force sensors for 10 seconds (to figure out the offset later).
- ☐ Stop measuring the force sensors.
- ☐ Unlock the cables from the participant and carefully place them, and the force sensors, in a safe spot.
- ☐ Unlock the participant from the BWS (the participant keeps on the harness).

- ☐ Check if all files are saved in the designated folders. (.csv, and .mp4)
- ☐ The participant will sit down on a chair, and the experimenter will take off the first design of both ankles. The doffing time is recorded with a stopwatch, by the experimenter.
  Attachment design round 2:.....
  Doffing time experimenter:.....
- ☐ Optional: notes. .....

- ☐ The experimenter will check both ankles for any sore spots or redness.
- ☐ Any redness/sore spots.

- ☐ The participant will fill in Questionnaire 1 (parts 1 and 2), evaluating the first design while resting for a moment.
- ☐ The experimenter checks the attachment design for damage.

After 15 minutes (since the stop of the treadmill) of rest for the participant (or longer/shorter depending on the preference of the participant), the experiment continues. (Minimum rest time is 5 minutes.)

- ☐ The participant sits on a chair while the experimenter puts on the second design on the participant's ankles. The donning time will be recorded with a stopwatch, by the experimenter and noted below.
  Attachment design round 3:.....
  Donning time experimenter:.....
- ☐ Optional: notes. .....

- ☐ Before entering the treadmill with the second design around both ankles, they are checked by the experimenter, for a safety check.

(a) Page five.

Part three of the experiment will begin, where the participant walks on the treadmill for another 20 minutes wearing the second design.

When cuff, measure the length from ground to bottom line of the cuff before and after walking for 20 min. on the treadmill.

CUFF HEIGHT BEFORE 20 MIN WALK: .....
CUFF HEIGHT AFTER 20 MIN WALK:.....

- ☐ Lock the participant back in the BWS.
- ☐ The participant will stand still in the middle (indicated by the marked points) of the treadmill and MATE will be attached to the design on both ankles.
- ☐ Check if the force sensors are attached to the left leg, and the tension hooks are on the right leg.
- ☐ Use the same cable lengths as in part 2.
- ☐ Check if the video recording is focused on the right frame. In the sagittal plane on the left side of the participant, are the following things still visible when the feet are ± 60 cm from the start place: Hip, knees, ankles, attachment designs.
- ☐ Check if the cables of the force sensor, and the ends of the elastic cables, are fasted to the participant's legs with Coban tape. Without taking away the working quality of the cables or influencing the participant's gait cycle.
- ☐ Ask if the participant is troubled by the cables. If yes, repeat the checkpoint above.
- ☐ Optional: notes: .....

- ☐ Ask if the participant is ready to start.
- ☐ Start measuring the force sensors, while the participant is still at a standstill at the (middle) place indicated by the markers.
- ☐ Measure force sensors for 10 seconds (to figure out the offset later).
- ☐ Stop measuring the force sensors.
- ☐ Ask if the participant is ready to start walking.
- ☐ Start the treadmill.
- ☐ Set walking speed to the preferred walking speed measured in part 1.
- ☐ Start the video and force sensors.
- ☐ Check if the video and force sensors are being recorded.
  The participant now walks on a treadmill.
- ☐ Tell the participant to look forward as much as possible.
- ☐ During the experiment keep checking if the pulleys are in the right spot and if the participant walks in the middle of the treadmill (indicated by the marked points).
- ☐ Optional: notes. .....

- ☐ Stop video and force sensor recording at minute 20 of the experiment (timing starts once the participant is walking in the preferred speed).
- ☐ Stop the treadmill right after.

(b) Page six.

- ☐ The participant will stand still in de middle of the treadmill (indicated by the markers).
- ☐ Start measuring the force sensors (while the participant is still at a standstill at the (middle) place indicated by the markers).
- ☐ Measure force sensors for 10 seconds (to figure out the offset later).
- ☐ Stop measuring the force sensors.
- ☐ Unlock the cables from the participant and carefully place them, and the force sensors, in a safe spot.
- ☐ Measure force sensors again for 10 seconds (to check if the null measurement values obtained in part one of the experiment are still the same).
- ☐ Unlock the participant from the BWS.

- ☐ Check if all files are saved in the designated folders. (.csv, and .mp4)
- ☐ The participant will sit down on a chair, and the experimenter will take off the second design of both ankles. The doffing time is recorded with a stopwatch, by the experimenter.
  Attachment design round 3:.....
  Doffing time experimenter:.....
- ☐ Optional: notes. .....

- ☐ The experimenter will check both ankles for any sore spots or redness.
- ☐ Any redness/sore spots.

- ☐ The participant will fill in Questionnaire 2 (parts 1 and 2), evaluating the second design while resting for a moment.
- ☐ The experimenter checks the attachment design for damage.

- ☐ The participant sits back on the chair to start the donning and doffing process of the first attachment design by themselves. The experimenter times the process and makes notes.
  Attachment design:.....
  Donning time participant:.....
- ☐ Optional: notes. .....

Doffing time participant:.....
- ☐ Optional: notes. .....

(c) Page seven.

- ☐ The participant takes a minute of rest.
- ☐ The participant starts the donning and doffing process of the second attachment design by themselves. The experimenter times the process and makes notes.
  Attachment design:.....
  Donning time participant:.....
- ☐ Optional: notes. .....

Doffing time participant:.....
- ☐ Optional: notes. .....

- ☐ The participant fills in part 3 in <u>both</u> questionnaires.
- ☐ Ask if the participant is okay, and has any questions.
- ☐ Optional: notes. .....

- ☐ This is the end of the experiment.

(d) Page eight.

Fig. 41: The experiment checklist completed for every participant.

Cleaning process

- ☐ Check if all video recordings (.mp4) and force sensor data (.csv) are saved in the correct folder.
- ☐ Check if the Python script is saved and closed correctly.
- ☐ Put the informed consent, the two questionnaires, and this checklist in a secured map.
- ☐ Careful store away the ZED mini camera and force sensors.
- ☐ Clean both designs with anti-bacterial wipes.
- ☐ Check both designs for damage.
- ☐ Remove cables from the frame, and check on damage.
- ☐ Store away cables and attachment designs securely.
- ☐ Store away the BWS harness, and check on damage.
- ☐ Restore cable lengths to start lengths:
  Cable front high: 0.69 m
  Cable front low: 1.29 m
  Cable back high: 0.97 m
  Cable back low: 1.69 m

At the end of the day

- ☐ Fill in the participant's details and notes from this checklist in the designated documents, and SAVE.
- ☐ Fill in the donning and doffing time in the 'donning/doffing' document, and SAVE!
- ☐ Fill in the answers of the questionnaire in the 'answers questionnaire' document, and SAVE!
- ☐ Give the signed informed consent paper of the participant to Laura.
- ☐ Put participant folder in 'RAW DATA' map.
- ☐ Make a back-up of computer!!!!

Fig. 42: The experiment checklist completed for every participant. Page nine.

APPENDIX G
INFORMED CONSENT FORM

INFORMED CONSENT FORM

You are being invited to participate in a research study titled [Minimally Actuated Tendon-based Exercise environment (MATE)]. This study is done by Belle Hopmans (Master student) from the TU Delft under Laura Marchal Crespo (Associate professor, CoR department, 3mE), Heike Vallery (Professor, department BMechE, 3mE), and Katherine Poggensee (Postdoctoral researcher, CoR department, 3mE).

The purpose of this research study is to investigate the forces acting on the legs during normal gait (walking) when using the MATE and two different attachment designs. The experiment will take approximately 90 minutes to complete. The data will be used for evaluating the MATE, writing a thesis, and patenting the device. We will be asking you to walk for 45 min in total (1x5min and 2x20 min) on a treadmill while always wearing a safety harness. First, you will walk 5 minutes without being attached to the MATE. Then, you will walk two sets of 20 minutes while being attached to the elastic cables from the MATE. For the first 20 min, you will wear and walk in the first attachment design. After a moment of rest and filling in a questionnaire, you start the second 20-minute walk in the other attachment design. This is again followed by filling in a questionnaire.

Your participation in this study is entirely voluntary and you can withdraw at any time. You are free to omit any questions. The collected data will be anonymized and as such cannot be removed after the experiment.

After filling in the survey, please provide your name, signature, and date on the indicated lines. Information details of the researchers are also included at the bottom of the form.

PLEASE TICK THE APPROPRIATE BOXES	Yes	No
A: GENERAL AGREEMENT – RESEARCH GOALS, PARTICPANT TASKS AND VOLUNTARY PARTICIPATION		
1. I have read and understood the study information dated 30/10/2023, or it has been read to me. I have been able to ask questions about the study and my questions have been answered to my satisfaction.	☐	☐
2. I consent voluntarily to be a participant in this study and understand that I can refuse to answer questions and I can withdraw from the study at any time, without having to give a reason.	☐	☐
3. I understand that taking part in the study involves:	☐	☐
• The information of this survey being stored locally, and I will be allowed to keep a copy. • Data on my gait cycle being collected with AR motion capture. • Video recording of the lower body, without audio.		
4. I understand that I will not be compensated for my participation.	☐	☐
5. I understand that the study will end after a single session, which is about 90 minutes. The full study will end in March 2024.		
B: POTENTIAL RISKS OF PARTICIPATING (INCLUDING DATA PROTECTION)		
6. I understand that taking part in the study involves the following risks. I understand that these will be mitigated by…	☐	☐
• Risk: Falling / Balance issues o Mitigation: Body weight support system, wearing a harness. • Risk: Hitting objects o Mitigation: Harness prevents too far movement, covers on moving objects, standby assistant, devices are validated for safety and/or CE certified. • Risk: Physical discomfort o Mitigation: Emergency stop provided. Can stop at any time, for any reason.		

Fig. 43: Page one. All participants were required to complete and sign the informed consent form before taking part in the experiment.

PLEASE TICK THE APPROPRIATE BOXES	Yes	No
7. I understand that taking part in the study also involves collecting specific personally identifiable information (PII) [information on this survey] and associated personally identifiable research data (PIRD) [gait cycle] with the potential risk of my identity being revealed [no additional risks determined].	☐	☐
8. I understand that the following steps will be taken to minimise the threat of a data breach and protect my identity in the event of such a breach [anonymized gait cycle data collection].	☐	☐
9. I understand that personal information collected about me that can identify me, such as [name], will not be shared beyond the study team.	☐	☐
C: RESEARCH PUBLICATION, DISSEMINATION AND APPLICATION		
10. I understand that after the research study the de-identified information I provide will be used for…	☐	☐
• Evaluating the MATE • Writing a thesis • Patenting the device		
11. I agree that my responses, views, or other input can be quoted anonymously in research outputs	☐	☐
D: (LONGTERM) DATA STORAGE, ACCESS AND REUSE		
12. I give permission for the de-identified [gait cycle data] that I provide to be archived in [a local database/storage] repository so it can be used for future research and learning.	☐	☐

Signatures

_____ _____ _____

Name of participant Signature Date

I, as researcher, have accurately read out the information sheet to the potential participant and, to the best of my ability, ensured that the participant understands to what they are freely consenting.

_____ _____ _____

Researcher name Signature Date

Study contact details for further information:

[Belle Hopmans, b.c.hopmans@student.tudelft.nl]

[Laura Marchal Crespo, L.MarchalCrespo@tudelft.nl]

Fig. 44: Page two. All participants were required to complete and sign the informed consent form before taking part in the experiment.

APPENDIX H
PYTHON SCRIPT USED FOR EXPERIMENTS

```
from threading import Thread
from scipy.spatial.transform import Rotation
import numpy as np
import pyzed.sl as sl
import serial
import csv
import cv2
import time
import os
import datetime

PORT            = 'COM3'
RESOLUTION      = 720
PRACTICE_TIME   = 5 #min
TRIAL_TIME      = 20 #min
ZEROTIME        = 10 #s

SHOW    = True # whether or not to show the capture live
FITTING = False # wheter or not to apply body fitting while tracking
BODY18 only
BODY    = sl.BODY_FORMAT.BODY_18
POINTS  = [1]+list(range(8,13+1))

FONT   = cv2.FONT_HERSHEY_SIMPLEX
FOURCC = cv2.VideoWriter_fourcc(*'mp4v')  # You can also use other codecs
like 'MJPG' or 'DIVX'

def serial_reader(ser):
    global last_received
    while True:
        try:
            line = ser.readline().decode('utf-8').strip()
            force_values = line.split(' ')
            force1, force2 = map(float, force_values)
            last_received = [force1, force2]
        except ValueError:
            last_received = [0,0]
            continue

def convert_time(t_seconds):
    # Convert seconds to timedelta
    delta = datetime.timedelta(seconds=t_seconds)

    # Extract minutes, seconds, and milliseconds
    hours = delta.seconds // 3600
    minutes = delta.seconds // 60
    seconds = delta.seconds % 60
    milliseconds = delta.microseconds // 1000

    # Display the time in the desired format
    if hours > 0:
        formatted_time =
f"{hours:02}:{minutes:02}:{seconds:02}:{milliseconds:03}"
    else:
        formatted_time = f"{minutes:02}:{seconds:02}:{milliseconds:03}"

    return formatted_time
```

```
def record_forces(fname, duration):
    print('Recording forces....')
    csv_file = open(fname, 'w', newline='')
    csv_writer = csv.writer(csv_file)
    csv_writer.writerow(['time', 'force_front', 'force_back'])
    t = 0.0
    start = time.time()

    while t<duration:
        t = time.time() - start
        force1, force2 = last_received
        csv_writer.writerow([t, force1, force2])
        time.sleep(1/50)

    csv_file.close()
    print("Done."), print()

#define csv header
body_header = []
for i in POINTS:
    body_header.extend([f'{i}_x', f'{i}_y', f'{i}_z', f'{i}_confidence'])
body_header.extend(['rot_x', 'rot_y', 'rot_z', 'rot_q'])
header = ['frame', 'fps', 'dt', 'overall_time', 'condition_time',
'force_front', 'force_back'] + body_header

# Initialize the serial connection
last_received = [0,0]

try:
    ser = serial.Serial(PORT, 9600)
    Thread(target=serial_reader, args=(ser,)).start()
except:
    print("Warning: Force sensor not connected")

# Wait for the Enter key
while True:
    try:
        participant = int(input("Enter participant number: "))  # Get
user input and convert it to an integer
        try:
            os.mkdir(str(participant))
            break
        except OSError as e:
            print(f'Error: {e}')
            break
    except ValueError:
        print("Invalid input. Please enter a valid number.")

print(), input(f'Press enter to start recording the pre-experiment
zeropoint for {ZEROTIME}s')
record_forces(str(participant)+f'/zeropoint_pre_experiment_{participant}.
csv', ZEROTIME)

print('Initializing ZED ...')
# Initialize ZED capture
zed = sl.Camera()

# Set configuration parameters
```

```
init_params = sl.InitParameters()
init_params.depth_mode = sl.DEPTH_MODE.NEURAL
init_params.coordinate_units = sl.UNIT.METER
# init_params.coordinate_system = sl.COORDINATE_SYSTEM.LEFT_HANDED_Y_UP
init_params.sdk_verbose = True

if RESOLUTION == 720:
    init_params.camera_resolution = sl.RESOLUTION.HD720 # Use HD720 opr
HD1200 video mode, depending on camera type.
    init_params.camera_fps = 30 # Set fps at 30
    FPS = 30
elif RESOLUTION == 1080:
    init_params.camera_resolution = sl.RESOLUTION.HD1080 # Use HD720 opr
HD1200 video mode, depending on camera type.
    init_params.camera_fps = 30 # Set fps at 30
    FPS = 30
elif RESOLUTION == 2160:
    init_params.camera_resolution = sl.RESOLUTION.HD2K # Use HD720 opr
HD1200 video mode, depending on camera type.
    init_params.camera_fps = 15 # Set fps at 30
    FPS = 15

# Open the camera
image = sl.Mat()
err = zed.open(init_params)
if (err != sl.ERROR_CODE.SUCCESS) :
    exit(-1)

# Define the Objects detection module parameters
print('Loading body tracking module ...')
body_params = sl.BodyTrackingParameters()
body_params.detection_model = sl.BODY_TRACKING_MODEL.HUMAN_BODY_ACCURATE
body_params.body_format = BODY
body_params.enable_tracking = True
body_params.image_sync = True
body_params.enable_segmentation = False
body_params.enable_body_fitting = FITTING
body_params.allow_reduced_precision_inference = False

# Object tracking requires the positional tracking module
if body_params.enable_tracking:
    positional_tracking_param = sl.PositionalTrackingParameters()
    positional_tracking_param.set_as_static = True
    positional_tracking_param.set_floor_as_origin = True
    zed.enable_positional_tracking(positional_tracking_param)

err = zed.enable_body_tracking(body_params)
if err != sl.ERROR_CODE.SUCCESS:
    print("Enable Body Tracking : "+repr(err)+". Exit program.")
    zed.close()
    exit()

bodies = sl.Bodies()
body_runtime_param = sl.BodyTrackingRuntimeParameters()
# For outdoor scene or long range, the confidence should be lowered to
avoid missing detections (~20-30)
# For indoor scene or closer range, a higher confidence limits the risk
of false positives and increase the precision (~50+)
```

```
body_runtime_param.detection_confidence_threshold = 40

### Start main loop  ----------------------------------------------------
----------------------------
trial = 'groundtruth'
start = time.time()
endtime = PRACTICE_TIME*60 #s

while True:
    if trial != 'groundtruth':
        print(), input(f'Press enter to start recording the pre-{trial}
pretension for {ZEROTIME}s')

record_forces(str(participant)+f'/pretension_pre_{trial}_{participant}.cs
v', ZEROTIME)

    print(f"Initializing condition {trial}'...")
    # Open a CSV file for writing
    csv_file =
open(str(participant)+f'/experiment_data_{trial}_{participant}.csv', 'w',
newline='')
    csv_writer = csv.writer(csv_file)
    csv_writer.writerow(header)

    # Get the frame width, height, and frames per second (fps) of the
input video
    width = int(RESOLUTION/9*16)
    height = int(RESOLUTION)

    # Define the codec and create a VideoWriter object
    out =
cv2.VideoWriter(str(participant)+f'/capture_{trial}_{participant}.mp4',
FOURCC, FPS, (width, height))

    recording = False
    t = 0.0
    trial_start = frame_time = time.time()
    frame_number = 1
    dt = 1/FPS

    print(), print("Press 'r' to start (or stop) recording")

    while t <= endtime:
        try:
            if zed.grab() == sl.ERROR_CODE.SUCCESS:
                T = round(frame_time-start,3)
                t = round(frame_time-trial_start,3)
                fps = int(1/dt)
                flattened_frame_data = [0]*18*3

                # read latest forces
                force1, force2 = last_received

                # Capture frame from webcam
                zed.retrieve_image(image, sl.VIEW.LEFT)
                frame = cv2.cvtColor(image.get_data(),
cv2.COLOR_RGBA2RGB)
```

```
                # detect bodies
                zed.retrieve_bodies(bodies, body_runtime_param)

                if bodies.is_new:
                    body_array = bodies.body_list
                    try:
                        first_body = body_array[0]

                        keypoint_3d = [first_body.keypoint[POINT] for
POINT in POINTS]
                        keypoint_2d = [first_body.keypoint_2d[POINT] for
POINT in POINTS]
                        keypoint_confidence  =
[first_body.keypoint_confidence[POINT] for POINT in POINTS]
                        global_body_orientation =
first_body.global_root_orientation
                        # print(f'Global Orientation:
{global_body_orientation}')

                        if BODY != sl.BODY_FORMAT.BODY_18:
                            keypoint_position_local =
[first_body.local_position_per_joint[POINT] for POINT in POINTS]
                            # print(f'Positon:
{keypoint_position_local}')
                            keypoint_orientation_local =
[first_body.local_orientation_per_joint[POINT] for POINT in POINTS]
                            # print(f'Orientation:
{keypoint_orientation_local}')

                        for idx, key in enumerate(keypoint_2d, start=0):
                            if POINTS[idx] > 9:
                                d = 5
                            else:
                                d = 2

                            # Draw circle and Add keypoint number as text
next to the circle
                            cv2.circle(frame, (int(key[0]), int(key[1])),
7, (0, 255, 0), -1)
                            cv2.putText(frame, str(POINTS[idx]),
(int(key[0]) -d, int(key[1]) +3), FONT, 0.25, (0, 0, 255), 1,
cv2.LINE_AA)

                        # flatten body 3d points
                        flattened_frame_data = []

                        for [key_3d, confidence] in zip(keypoint_3d,
keypoint_confidence):
                            flattened_frame_data.extend(key_3d)
                            flattened_frame_data.append(confidence)

flattened_frame_data.extend(global_body_orientation)

                    except Exception as err:
                        print(err)
                        body_array = []
```

```
                # Display the forces in the top left corner of the video
frames
                cv2.putText(frame, f'FPS: {fps}',
(10, 20), FONT, 0.4, (255, 0, 0), 1, cv2.LINE_AA)
                cv2.putText(frame, f'Frame: {frame_number}',
(10, 40), FONT, 0.4, (255, 0, 0), 1, cv2.LINE_AA)
                cv2.putText(frame, f'Time: {convert_time(T)}',
(10, 60), FONT, 0.4, (255, 0, 0), 1, cv2.LINE_AA)
                cv2.putText(frame, f'Condition endtime:
{convert_time(endtime)}',  (10, 80), FONT, 0.4, (255, 0, 0), 1,
cv2.LINE_AA)
                cv2.putText(frame, f'Condition runtime:
{convert_time(t)}',        (10,100), FONT, 0.4, (255, 0, 0), 1,
cv2.LINE_AA)
                cv2.putText(frame, f'Force front: {force1} N',
(10,120), FONT, 0.4, (255, 0, 0), 1, cv2.LINE_AA)
                cv2.putText(frame, f'Force back: {force2} N',
(10,140), FONT, 0.4, (255, 0, 0), 1, cv2.LINE_AA)

                if recording:
                    print(f'Frame {frame_number}: FPS: {fps}, Front:
{force1} N, Back: {force2} N, Person: {len(body_array)}')

                cv2.imshow('frame', frame)

                if cv2.waitKey(1) & 0xFF == ord('r'):
                    if not recording:
                        print(f'Recording condition {trial} ...')
                        recording = True
                    elif recording:
                        print(f'Stopping condition {trial}...')
                        break
                elif cv2.waitKey(1) & 0xFF == ord('q'):
                    break

                # Write the frame to the output video file and data to
csv file
                if recording:
                    csv_writer.writerow([frame_number, fps, dt, T, t,
force1, force2] + flattened_frame_data)
                    out.write(frame)
                    frame_number += 1
                else:
                    trial_start = time.time()

                dt = time.time()-frame_time
                frame_time = time.time()

        except KeyboardInterrupt:
            print(f'Stopping condition {trial} ...')
            break

    csv_file.close()
    out.release()
    cv2.destroyAllWindows()
    endtime = TRIAL_TIME*60 #s

    if trial != 'groundtruth':
```

```
        print(), input(f'Press enter to start recording the post-{trial}
pretension for {ZEROTIME}s')

record_forces(str(participant)+f'/pretension_post_{trial}_{participant}.c
sv', ZEROTIME)

    while True: # Ask the user whether to continue or quit
        try:
            print()
            user_input = str(input("Type the following condition
(a=cuff,b=bandage) to continue, or type 'q' to stop the experiment "))
            if user_input == 'q':
                print(), input(f'Press enter to start recording the post-
experiment zeropoint for {ZEROTIME}s')

record_forces(str(participant)+f'/zeropoint_post_experiment_{participant}
.csv', ZEROTIME)
                ser.close()
                zed.close()
                exit()
            elif user_input == 'a' and trial != "cuff":
                trial = "cuff"
                break
            elif user_input == 'b' and trial != "bandage":
                trial = "bandage"
                break
            else:
                print("Invalid input. Either condition already played, or
invalid character.")

        except ValueError:
            print("Invalid input. Please enter a valid number.")

        except KeyboardInterrupt:
            print('Stopping Experiment ...')
            ser.close()
            zed.close()
            exit()

        except Exception as e:
            ser.close()
            zed.close()
            exit()
```

APPENDIX I
DETAILED FORCE CALCULATIONS DOCUMENT

Research background

In this document, a better understanding of the effect of MATE on the participant will be created. Cable forces are calculated in different scenarios, which is done by implementing the corresponding ankle positions in the 'PreTensionCablesWITHrighthandrule.m' Matlab script. Normal gait cycle scenarios, critical, and worst-case scenarios are considered. With the obtained information, the forces on the attachments can be calculated, and safety measurements can be put in place to ensure user safety. This document also explains the decisions made during the calculations and how they are implemented in the Matlab script.

General information

The x, y, and z-axis belong to the following orientations. A visualisation is shown below in Figure 1.

- The x-axis is the frontal/coronal plane, where the 0 point is in the middle of the treadmill and the positive direction points to the anterior (the front of the body).
- The y-axis is the sagittal plane, where 0 is again the middle of the treadmill (matching the zero point of the x-axis), and the positive direction is to the left side of the body.
- The z-axis is the transversal plane, where 0 is in the middle of the treadmill (matching the x and y-axis) and is placed on the ground. The positive direction is going up.

The human is always assumed to be in the middle of the treadmill. However, the person's legs will move in the x, y, and z directions. Here, the human ankles indicate where the legs are and in what part of the gait cycle the person is. Also, MATE will be attached to the human at/close to the ankle joint.

The used coordinate system indicates that when a direction or force is a negative number in the x-element, the legs are moved/pulled backwards. When the y-axis has a negative number, the legs are moved/pulled to the right. When the z-element is negative, the legs are moved/pulled downwards.

Furthermore, the ankle is indicated with one specific point in the middle. This assumption is made because of the differences in thickness between people's ankles. Also, by taking the middle point of the ankle, a bigger extension u is considered when doing calculations, and this will result in a bigger cable force than would be in reality. Making sure the attachment can hold the bigger force will increase the user's safety.

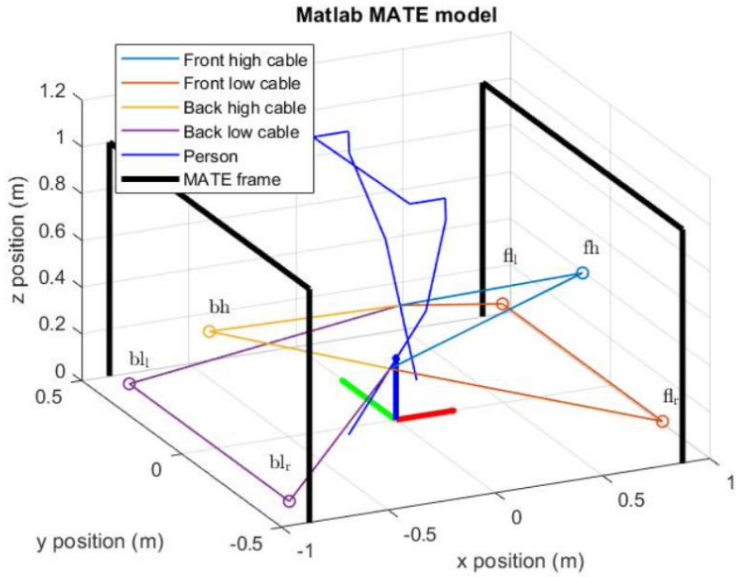


Figure 1: MATE with the coordinate system. This is Kieft's paper, figure 2 [1].

1. Frame coordinates in Matlab: lines 1-47

The positions of the pulleys are shown in Figure 1. In Matlab, the positions of the pulleys are marked as follows:

- Figure 2 shows the x, y, and z coordinates of the back high (bh), back low right (blr), and back low left (bll) pulleys.
- Figure 3 shows the x, y, and z coordinates of the front high (fh), front low right (flr), and front low left (fll) pulleys.

```
% BACK-----------------

% Back high (BH)
BHx = -0.9  % meters, x position BH
BHy = 0     % meters, y position BH
BHz = 0.5   % meters, z position BH

BH = [BHx BHy BHz];

% Back low, right (BLr)
BLrx = -0.9  % meters, x position BLr
BLry = -0.4   % meters, y position BLr
BLrz = 0.1   % meters, z position BLr

BLr = [BLrx BLry BLrz];

% Back low, left (BLl)
BLlx = -0.9   % meters, x position BLl
BLly = 0.4   % meters, y position BLl
BLlz = 0.1    % meters, z position BLl

BLl = [BLlx BLly BLlz];
```

Figure 2: Matlab back frame positions

```
% FRONT-----------------

% Front High (FH)
FHx = 0.9   % meters, x position FH
FHy = 0     % meters, y position FH
FHz = 0.5   % meters, z position FH

FH = [FHx FHy FHz];

%Front low, right (FLr)
FLrx = 0.9   % meters, x position FLr
FLry = -0.4   % meters, y position FLr
FLrz = 0.13  % meters, z position FLr

FLr = [FLrx FLry FLrz];

% Front low, left (FLl)
FLlx = 0.9    % meters, x position FLl
FLly = 0.4   % meters, y position FLl
FLlz = 0.13   % meters, z position FLl

FLl = [FLlx FLly FLlz];
```

Figure 3: Matlab front frame positions

2. Ankle positions: lines 50-182

Different gait cycle scenarios are used to better understand the cables' effect on the attachments/human legs in those scenarios. As mentioned in the subchapter 'research background', different situations are considered in this script (our shortcuts): normal gait cycle, critical, and worst-case scenarios. The inclusion of stroke gait in these scenarios was omitted because of the differences between stroke patients and their gait cycles [2] [3]. However, the used worst-case scenarios are designed to be the actual worst-case scenarios, so all different stroke gait cycles fall within the boundaries of these worst-case cases (considering the stroke gait of Balaban [3]). The shortcuts to these different scenarios can also be used for quick checks on attachment safety and feedback on how much 'help' the participant receives from the cables to perform their gait cycle.

At the top of the Matlab section, "Positions ankles general," is a field to experiment with different positions of both ankles without changing the prepared scenarios (our shortcuts); see Figure 4.

```
%% Positions ankles general

% Right leg, position ankle
AnkleRx = 0      % meters
AnkleRy = -0.15  % meters
AnkleRz = 0.1    % meters

AnkleR = [AnkleRx AnkleRy AnkleRz];

% Left leg, position ankle
AnkleLx = 0      % meters
AnkleLy = 0.15   % meters
AnkleLz = 0.1    % meters

AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 4: experiment with different ankle positions.

The following scenarios, the shortcuts, are pre-made. When you want to check one of these shortcuts, first 'comment' the currently used ankle positions, then 'uncomment' the shortcut scenario you want to check.

The shortcuts for the ultimate scenarios are shown in Figures 5 to 9.

```
% ------------------------------------------------------------
% ULTIMATE SCENARIO'S USING MATE

%   1) Both feet are placed in the back of the treadmill
% Right leg, position ankle
% AnkleRx = -0.85      % meters
% AnkleRy = -0.4      % meters
% AnkleRz = 0.1      % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = -0.85      % meters
% AnkleLy = 0.4     % meters
% AnkleLz = 0.1      % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 5: ultimate scenario 1.

```
%   2) Both feet are place in the front of the treadmill
% Right leg, position ankle
% AnkleRx = 0.85      % meters
% AnkleRy = -0.4      % meters
% AnkleRz = 0.1      % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = 0.85      % meters
% AnkleLy = 0.4      % meters
% AnkleLz = 0.1      % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 6: ultimate scenario 2.

```
%   3) Both feet are place in the middle (wide)
% of the treadmill, both feet are on the ground
% Right leg, position ankle
% AnkleRx = 0     % meters
% AnkleRy = -0.4     % meters
% AnkleRz = 0.1     % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = 0     % meters
% AnkleLy = 0.4     % meters
% AnkleLz = 0.1     % meters
% %
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 7: ultimate scenario 3.

```
%   4) Both feet are place in the middle (wide) of the
% treadmill (right leg is on the ground and left leg is
% in the air, trying to get the ultimate stiff leg scenario)
% % Right leg, position ankle
% AnkleRx = 0     % meters
% AnkleRy = -0.4     % meters
% AnkleRz = 0.1     % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = 0     % meters
% AnkleLy = 0.4     % meters
% AnkleLz = 0.2     % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 8: ultimate scenario 4.

```
%   5) Stiff leg (normal right leg, mid-stance)
% (stiff left leg, mid-swing, depence on height
% of the person's legs)
% % Right leg, position ankle
% AnkleRx = 0     % meters
% AnkleRy = -0.1     % meters
% AnkleRz = 0.1     % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = 0     % meters
% AnkleLy = 0.4     % meters
% AnkleLz = 0.2     % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 9: ultimate scenario 5.

Ultimate scenario 1, where both ankles are in the back of the treadmill, is chosen because of the possibility of failing to perform the gait cycle and falling during walking. The participant is held upright with the help of the body support system. However, the legs can still be pulled backwards because of the direction of a treadmill in use.
The ankle coordinates in this situation, shown in Figure 5, have been established as follows:

- The x-position of the pulleys in the back is -0.9m. The frame that holds to pulleys surrounds the treadmill, so in an ultimate scenario, the x-positions of the ankles can't be the same as the x-positions of the pulleys. Therefore, it is assumed that the x-positions of both ankles in this scenario are -0,85m.
- The y-positions are assumed to be most extreme at the sides of the treadmill. At positions -0.4m (right foot) and 0.4m (left foot), the forces in the cables are higher than when the feet are placed at -0.1m and 0.1m, respectively.
- Assuming both feet are still placed on the treadmill in this scenario, the ankle height is 0.1m (z-position of both feet). The information on the height of the ankle being 0.1m when the foot is placed in the ground is obtained by eyeballing this situation in Figure 10 and considering the information in Figure 11, where the average instep height of men and women is shown [4]. Adding the height of a shoe sole to this instep height and around a centimetre to get to the ankle joint, it can be assumed that this final height will be the 0.1m mentioned earlier.

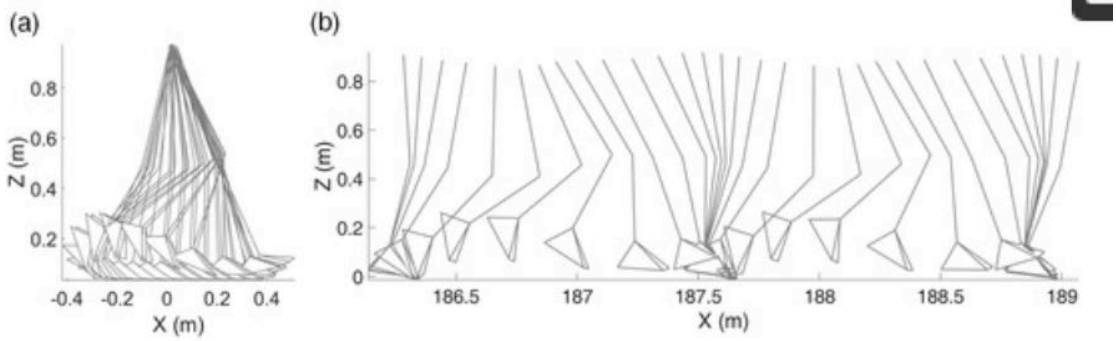


Figure 10: Information on ankle positions in different scenarios in the gait cycle [5].

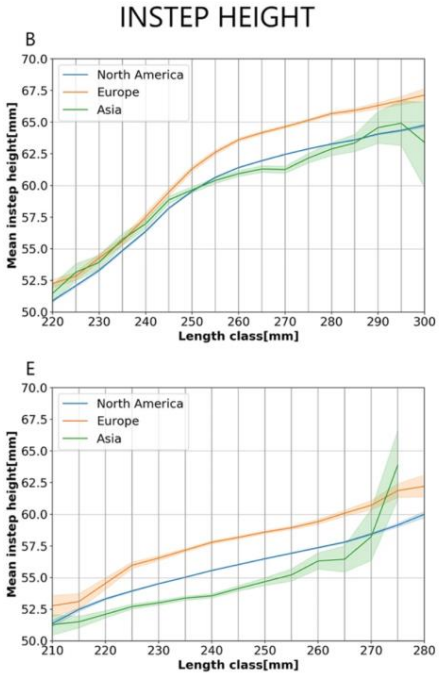


Figure 11: Top graph: mean instep height male. Bottom graph: mean instep height female.

Ultimate scenario 2, see Figure 6, is similar to ultimate scenario 1. The only difference is that both feet are now placed in the front of the treadmill. This scenario can happen when the participant walks too fast and bumps into the front of the treadmill. In this situation, the ankle coordinates (and the reasoning behind the positions) are close to equal to the ankle coordinates in Figure 5. The only difference is that the x-position in this situation is 0.85m for both ankles and not -0.85m.

Ultimate scenario 3, where both feet are placed on the outer sides of the treadmill, could happen when the participant gets scared and steps on the sides of the treadmill, away from the moving part. The ankle coordinates in this situation, shown in Figure 7, are selected as follows.

- The x-position for both feet is 0m because it's assumed that the participant is standing in the middle of the treadmill.

- The y-position is assumed to be 0.4m and -0.4m, where both feet can be placed on the outer sides of the treadmill. (The width of the moving band of the treadmill is 67 cm, and the full-width treadmill is 90 cm.)
- Because both feet are assumed to be on the ground in this scenario, the z-position of the ankles is again 0.1m (explanation in ultimate scenario 1).

Ultimate scenario 4 in Figure 8 and ultimate scenario 5 in Figure 9 are two situations that refer to walking with a stiff leg. In **ultimate scenario 4**, the most extreme version of the two is shown. Both feet are located on the outer sides in the middle of the treadmill (x-coordinates are 0m, and the y-coordinates are -0.4 (right ankle) and 0.4m (left ankle)). One foot is on the ground, and the other is in the air, as would be the case during mid-swing and mid-stance state. The chosen z-positions for the ankles are 0.1m (foot standing on the ground), and for the swinging leg, it's assumed to be lifted from the ground by 0.2m (swinging foot).

Ultimate Scenario 5 represents a less extreme version of a stiff-leg scenario. Again, the stance foot is in the middle of the treadmill (x-coordinate is 0m). However, this time, the leg and ankle are placed under the participant instead of wide on the outer sides of the treadmill. A normal step width would be between 3-10 cm (see normal scenario 1) but given this is still a 'stiff leg scenario' and that step width increases with instability [6], the y-coordinate is assumed to be -0.1m. The z-coordinate for this foot remains the same at 0.1m. The swing leg has the same coordinates as in ultimate scenario 4.

The shortcuts for the normal scenarios are shown in Figures 12 and 13.

```
% ------------------------------------------------------
% NORMAL SCENARIO'S USING MATE

% 1) Right leg in heel-strike, left leg in toe-off
% (not the situation in real life, but the combination
% of a heel strike and a toe-off is the most extreem
% version of making big steps.)
% Right leg, position ankle
% AnkleRx = 0.36  % meters
% AnkleRy = -0.07 % meters
% AnkleRz = 0.1   % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = -0.36  % meters
% AnkleLy = 0.07   % meters
% AnkleLz = 0.15   % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 12: normal scenario 1.

```
% 2) Right leg in mid-stance, left leg in mid-swing
% % Right leg, position ankle
% AnkleRx = 0     % meters
% AnkleRy = -0.07  % meters
% AnkleRz = 0.1   % meters
%
% AnkleR = [AnkleRx AnkleRy AnkleRz];
%
% % Left leg, position ankle
% AnkleLx = 0     % meters
% AnkleLy = 0.05 % meters
% AnkleLz = 0.2   % meters
%
% AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 13: normal scenario 2.

Normal scenario 1, see Figure 12, represents the part in the gait cycle where one foot is in toe-off and the other foot in heel strike. In reality, the first moment of heel strike and the last moment of toe-off don't happen simultaneously, as this is an overlapping process. However, when taking the most extreme version of this 'normal' scenario, it is assumed these moments do happen simultaneously. Therefore, when the right foot is in heel strike and the left foot in toe-off, the coordinates are:
The x-coordinate for heel strike is 0.36m and for toe-off -0.36m. The step length depends on the length (of the legs) of a person and is, therefore, variable. Also, strength, balance and

muscle fatigue play an important role [5] [7]. For now, a step length of 72 cm is assumed to be the case [8], which compares to the information in Figure 10 at the heel strike and toe-off ankle positions.

- The average step width normally lies between 3-10cm [7] [8] [9]. However, also the step width is influenced by instability, causing a larger step width in older age groups [7][9]. The y-positions used are a combination of the information above and the y-results in figure 14, resulting in -0.07m for the heel strike, and 0.07m for the toe-off.
- With the help of Figure 10, the z-coordinates are assumed to be 0.1m in heel strike position and 0.15m in toe-off.

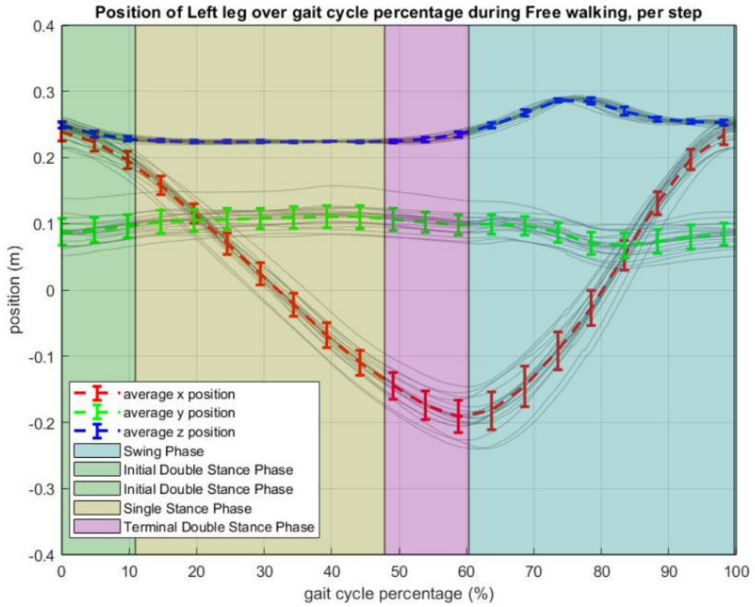


Figure 14: average x,y,z-coordinates Left leg measured by Kieft (figure 20 in Kieft [1]).

The attachments used in Kieft's study [1], were connected to the legs when obtaining these results, and not to the ankles. Therefore, only the y-coordinates are useful for the scenarios mentioned in this document.

Normal scenario 2, Figure 13, represents another part of the gait cycle where the right leg is in mid-stance, and the left is in mid-swing. The coordinates in this position are:

- The x-coordinates for both feet are assumed to be 0m because both feet are in the 'mid'-point of their movement. Also, Figure 10 supports this assumption. (Although it is mentioned that Figure 14 is only used for the y-position of the ankles, it is clear that in mid-swing and mid-stance, the position of the studied leg is at 0m.)
- The y-coordinates are -0.07m in mid-stance and 0.05m in mid-swing. The difference in position is caused by the slight decrease in y-position during mid-swing, shown in Figure 14.
- Finally, the z-position in mid-stance is assumed to be 0.1m (see Figure 10 and the explanation in Ultimate Scenario 1), and the z-position for mid-swing is set at 0.2m (Figure 10).

3. Directional vector and angles: from the ankle to the frame

Directional vector: lines 185-293

Now that the positions of the ankles are known, the directional vector from the ankle to the pulleys of the frame can be calculated. Here, the ankle is the starting point, and the pulley on the frame is the terminal point. Also, once the directional vector is known, the corresponding directional angles can be obtained. This chapter provides some calculations for the right ankle to the frame. In Appendix A, the rest of the calculations for the right ankle can be found, and the left ankle is subsequently addressed.

The following formulas (1-3) were used in determining the directional vectors. This example is focused on the directional vector from the right ankle (AnkleR) to the pulley on the front high frame (FH). The values corresponding to FH and AnkleR can be found in Chapters 1 and 2, respectively.

$$FH = Position_{FH} = [FH_x\ FH_y\ FH_z] \quad (1)$$

$$Ankle_R = Position_{AnkleR} = [AnkleR_x\ \ AnkleR_y\ AnkleR_z] \quad (2)$$

$$Directional\ vector_{AnkleR\ to\ FH} = [FH - Ankle_R] \quad (3)$$

The first formula in Figure 15 represents the example above. It also shows all other directional vectors from the right ankle to the matching pulleys.

```
% RIGHT ankle to frame (directional vector: in x, y and z direction)------
RightAnkle_to_FrameFH = [FH - AnkleR]     % meters
RightAnkle_to_FrameFLr = [FLr - AnkleR]   % meters

RightAnkle_to_FrameBH = [BH - AnkleR]     % meters
RightAnkle_to_FrameBLr = [BLr - AnkleR]   % meters
```

Figure 15: directional vectors of the right ankle to the frame (in Matlab).

Directional angles: lines 195-297

The directional angles corresponding to the directional vectors shown above are obtained as follows.

- Alpha: the angle formed between the directional vector and the positive x-axis.
- Beta: the angle formed between the directional vector and the positive y-axis.
- Gamma: the angle formed between the directional vector and the positive z-axis.

The unit vectors corresponding to the positive axis are:

- For the x-axis: [1 0 0]
- For the y-axis: [0 1 0]
- For the z-axis: [0 0 1]

Finding alpha, beta and gamma is done with formulas 4, 5, and 6. The top of the fraction shows the dot product between the directional vector and a unit vector, and the bottom part shows the multiplication of the magnitudes of both vectors. In the following formulas:
v = directional vector
i = unit vector for the positive x-axis
j = unit vector for the positive y-axis
k = unit vector for the positive z-axis

$$alpha = cos^{-1}\left(\frac{v \cdot i}{|v||i|}\right) \quad (4)$$

$$beta = cos^{-1}\left(\frac{v \cdot j}{|v||j|}\right) \quad (5)$$

$$gamma = cos^{-1}\left(\frac{v \cdot k}{|v||k|}\right) \quad (6)$$

As mentioned in Figure 15, the directional vector from the right ankle to the pulley on the front high frame is RightAnkle_to_FrameFH. The example continues in Figure 16, where this directional vector's directional angles (in degrees) are obtained.

```
% Right ankle to frame:
% Front High
alpha_R_FH = acosd(dot(RightAnkle_to_FrameFH,i)/(norm(RightAnkle_to_FrameFH)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_R_FH = acosd(dot(RightAnkle_to_FrameFH,j)/(norm(RightAnkle_to_FrameFH)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_R_FH = acosd(dot(RightAnkle_to_FrameFH,k)/(norm(RightAnkle_to_FrameFH)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_R_FH_degrees = [alpha_R_FH   beta_R_FH   gamma_R_FH]
```

Figure 16: directional angles of the right ankle to the pulley on the front high frame (in Matlab).

An overview of all the directional angles of the right ankle to the frame, as well as all directional vectors and angles of the left ankle to the frame, can be found in Appendix A.

Finally, the magnitudes of the directional vectors of the right ankle to the frame are mentioned in the Matlab script (see Figure 17). The magnitude equals the absolute distance between the ankle and the corresponding pulley on the frame. These absolute distances are also used later in the Matlab script to calculate the cable extensions. The absolute distances of the left ankle to the corresponding pulleys on the frame can be found in Appendix A.

```
% RIGHT ankle to frame (absolute distance)--------------------------------
DistanceFH_RightAnkle = norm([FH - AnkleR])    % meters
DistanceFLr_RightAnkle = norm([FLr - AnkleR])  % meters

DistanceBH_RightAnkle = norm([BH - AnkleR])    % meters
DistanceBLr_RightAnkle = norm([BLr - AnkleR])  % meters
```

Figure 17: Absolute distance from the right ankle to the corresponding pulleys on the frame (in Matlab).

4. Cable force

Cable stiffness (C)

The forces in all four cables are obtained with the formula F = C * u. The force in a cable is represented by F, the cable stiffness = C, and the cable extension = u.

The cable stiffness is 160 N/m [1]. The previous master's student working on the MATE calculated this stiffness, and the same set of elastic cables is used for this experiment.

Cable extension (u)

The extension of the cables depends on the position of the right and left ankle and also on the unstretched length of the cable. The unstretched length of the cable correlates to the preferred pretension in the cable. So, before anything can be said about the extension of the cables and the matching cable forces, the pretension needs to be obtained.

4.1 Pretension

Obtaining the cable pretension

The pretension in the cables is obtained in script 'nieuwWITHrighthandrule.m' lines 315-375.

The pretension in the cables is measured while a participant is standing in the middle of the treadmill. So, the x-coordinates of both feet are 0m, and the z-coordinates are 0.1m, as explained in Chapter 2. The y-coordinates depend on the size of the participant, as the feet should be a shoulder-width apart when maintaining good standing posture [10]. The average shoulder width for women is 0.36m, and 0.41m for men [11]. For now, a shoulder width of 0.4 cm is used in the calculations, so the y-coordinates are -0.2m for the right ankle and 0.2m for the left ankle. The Matlab coordinates are shown in figure 18.

```
% 6) Standstill to calculate the pre-tension force (in the 4 cables) for
% the PreTensionCalblesWITHrighthandrule script!
% Right leg, position ankle
AnkleRx = 0     % meters
AnkleRy = -0.2  % meters
AnkleRz = 0.1   % meters

AnkleR = [AnkleRx AnkleRy AnkleRz];

% Left leg, position ankle
AnkleLx = 0     % meters
AnkleLy = 0.2 % meters
AnkleLz = 0.1   % meters

AnkleL = [AnkleLx AnkleLy AnkleLz]
```

Figure 18: Ankle position participant when at a standstill (in Matlab).

Additionally, the unstretched cable lengths are obtained from Table 2 (overall minimum in meters) from the master thesis of the previous master's student working on the MATE; see Figure 19 [1].

These unstretched/original cable lengths are:
OriginalCableLengthFH = 1.9223 m
OriginalCableLengthFL = 1.8371 m
OriginalCableLengthBH = 2.3203 m
OriginalCableLengthBL = 1.8371 m

Here, FH is short for front high, FL is front low, BH is back high, and BL is back low. Figure 1 provides a visual overview of these cables.

	subject	tendon			
		tfu	tfl	tru	trl
minimum length (m)	subject 1	1.9223	1.8371	2.3203	1.8371
	subject 2	1.9672	1.8840	2.3576	1.8840
	subject 3	1.9603	1.8769	2.3519	1.8769
	subject 4	1.9478	1.8638	2.3415	1.8638
	subject 5	1.9591	1.8756	2.3510	1.8756
	subject 6	1.9345	1.8499	2.3304	1.8499
maximum length (m)	subject 1	1.9548	1.8711	2.3474	1.8711
	subject 2	1.9921	1.9100	2.3784	1.9100
	subject 3	1.9971	1.9153	2.3827	1.9153
	subject 4	1.9846	1.9022	2.3721	1.9022
	subject 5	1.9855	1.9032	2.3729	1.9032
	subject 6	1.9545	1.8707	2.3470	1.8707
overal minimum	- (1)	1.9223	1.8371	2.3203	1.8371
overal maximum	- (3)	1.9971	1.9153	2.3827	1.9153
difference	-	0.0748	0.0782	0.0624	0.0782

Figure 19: Kieft's unstretched cable lengths [1].

The cable extension is obtained by measuring the distance from the right ankle and left ankle to a particular pulley (see figures 17, 18 and AC) minus the original length of the cable corresponding to that pulley. See Figure 20 for the formulas in Matlab.

A part of the front low cable and back low cable is between the left and right pulley. In Matlab, these lengths are mentioned as follows. For the numbers, see Figures 2 and 3.
CableLengthFL_middlepart = abs(FLry) + abs(FLly)
CableLengthBL_middlepart = abs(BLry) + abs(BLly)

```
% General extension (u) cable during use MATE (in meters)
u_cableFH = (DistanceFH_RightAnkle + DistanceFH_LeftAnkle) - OriginalCableLengthFH
u_cableFL = (DistanceFLr_RightAnkle + DistanceFLl_LeftAnkle + CableLengthFL_middlepart) - OriginalCableLengthFL

u_cableBH = (DistanceBH_RightAnkle + DistanceBH_LeftAnkle) - OriginalCableLengthBH
u_cableBL = (DistanceBLr_RightAnkle + DistanceBLl_LeftAnkle + CableLengthBL_middlepart) - OriginalCableLengthBL
```

Figure 20: cable extension (obtaining the pretension).

However, if the original cable length is longer than the distance between the pulley and both ankles, the cable extension becomes a negative number. This means there is no tension in the cable, and no tension means no cable force. So, whenever a negative extension is reached, the extension becomes 0. See Figure 21 for the formulas in Matlab.

```
u_cableFH = max(u_cableFH,0)
u_cableFL = max(u_cableFL,0)
u_cableBH = max(u_cableBH,0)
u_cableBL = max(u_cableBL,0)
```

Figure 21: Negative cable extension becomes zero.

Finally, the corresponding cable tension is calculated using the formula F = C * u (see Figure 22). As the last master's student working on the MATE did not define the pretension, these forces will be assumed to be the pretensions in the cables and used in the rest of the calculations.

Force_FH = 14.0280 Newton
Force_FL = 129.2456 Newton
Force_BH = 0 Newton
Force_BL = 129.0894 Newton

```
% F = C * u

% GENERAL cable forces (in Newton)
Force_FH = C * u_cableFH      % Force in the Front high cable
Force_FL = C * u_cableFL      % Force in the Front low cable
Force_BH = C * u_cableBH      % Force in the Back high cable
Force_BL = C * u_cableBL      % Force in the Back low cable
```

Figure 22: obtaining the cable pretension.

Obtaining cable length with pretension: lines 300-414

The rest of the document explains and elaborates on the remaining part of the script 'PreTensionCablesWITHrighthandrule'.

In the subchapter above, the pretension in the cables is obtained. However, it is unknown what the perfect pretension will be. In this subchapter, cable extension formulas will be created that depend on the pretension. Also, because the pretension needs to be the same for every person, and not every participant has the same width between their feet during standstill, the cable lengths keep changing.

The formulas in Figure 23 can be filled in with the correct ankle positions per participant during a standstill. The numbers shown in the figure are inspired by the average shoulder-width of men and women mentioned above.

```
% The pre-tenion is measured when a person is standing in the middle of the
% treadmill. The width of the ankles can be changed here:
% Right leg, position ankle
AnkleRx_atStandStill = 0     % meters
AnkleRy_atStandStill = -0.2  % meters
AnkleRz_atStandStill = 0.1   % meters
AnkleR_atStandStill = [AnkleRx_atStandStill   AnkleRy_atStandStill     AnkleRz_atStandStill];

% Left leg, position ankle
AnkleLx_atStandStill = 0     % meters
AnkleLy_atStandStill = 0.2 % meters
AnkleLz_atStandStill = 0.1   % meters
AnkleL_atStandStill = [AnkleLx_atStandStill   AnkleLy_atStandStill      AnkleLz_atStandStill]
```

Figure 23: Ankle positions during standstill.

The absolute distance from the ankles to the corresponding pulleys is measured using the formulas in Figure 24. When the length of the cable is equal to the sum of these absolute distances, there would be no extension, so no tension, in the cables (see Figure 25).

```
% RIGHT ankle to frame (absolute distance)---------------------------------
DistanceFH_RightAnkle_atStandStill = norm([FH - AnkleR_atStandStill])    % meters
DistanceFLr_RightAnkle_atStandStill = norm([FLr - AnkleR_atStandStill])  % meters

DistanceBH_RightAnkle_atStandStill = norm([BH - AnkleR_atStandStill])    % meters
DistanceBLr_RightAnkle_atStandStill = norm([BLr - AnkleR_atStandStill])  % meters

% LEFT ankle to frame (absolute distance)----------------------------------
DistanceFH_LeftAnkle_atStandStill = norm([FH - AnkleL_atStandStill])     % meters
DistanceFLl_LeftAnkle_atStandStill = norm([FLl - AnkleL_atStandStill])   % meters

DistanceBH_LeftAnkle_atStandStill = norm([BH - AnkleL_atStandStill])     % meters
DistanceBLl_LeftAnkle_atStandStill = norm([BLl - AnkleL_atStandStill])   % meters
```

Figure 24: absolute distance between ankles and pulleys during standstill.

```
% When a cable has an equal length as the summ of the distance between the trolley and the feet, there would be no tension.
% So,
lengthFH_NOtension = (DistanceFH_RightAnkle_atStandStill + DistanceFH_LeftAnkle_atStandStill)
lengthFL_NOtension = (DistanceFLr_RightAnkle_atStandStill + DistanceFLl_LeftAnkle_atStandStill + CableLengthFL_middlepart)
lengthBH_NOtension = (DistanceBH_RightAnkle_atStandStill + DistanceBH_LeftAnkle_atStandStill)
lengthBL_NOtension = (DistanceBLr_RightAnkle_atStandStill + DistanceBLl_LeftAnkle_atStandStill + CableLengthBL_middlepart)
```

Figure 25: Cable lengths when there is no tension.

However, in the experiment set-up it is not only the cable that is in the line between the pulley and the ankles. There are also three carabiners, a force sensor, and a tension hook. In

figure 26, the formulas are shown that incorporate all items from the experiment set-up. Either these formulas are used in the script, or the formulas mentioned in figure 25. The ones that are not used, are commented.

```
length_tensionhook = 0.07 % meters
length_forcesensor = 0.10 % neters
length_carabiner_small = 0.05 % meters
length_carabiner_big = 0.06 % meters
%
% % So, then the NO tension cables lenghts would be:
lengthFH_NOtension = (DistanceFH_RightAnkle_atStandStill + DistanceFH_LeftAnkle_atStandStill) - length_tensionhook
        - length_forcesensor - (2* length_carabiner_small) - length_carabiner_big
lengthFL_NOtension = (DistanceFLr_RightAnkle_atStandStill + DistanceFLl_LeftAnkle_atStandStill + CableLengthFL_middlepart)
        - length_tensionhook - length_forcesensor - (2* length_carabiner_small) - length_carabiner_big
lengthBH_NOtension = (DistanceBH_RightAnkle_atStandStill + DistanceBH_LeftAnkle_atStandStill) - length_tensionhook
        - length_forcesensor - (2* length_carabiner_small) - length_carabiner_big
lengthBL_NOtension = (DistanceBLr_RightAnkle_atStandStill + DistanceBLl_LeftAnkle_atStandStill + CableLengthBL_middlepart)
        - length_tensionhook - length_forcesensor - (2* length_carabiner_small) - length_carabiner_big
```

Figure 26: Cable lengths when there is no tension. In the experiment setup.

Now that the lengths of the cables without pretension are known, the extension in the cable that causes the pretension needs to be obtained. In subchapter 4.1, the preferred pretension was established and will be addressed as shown in figure 27 in the rest of the document. As mentioned above, the ideal pretension is still unknown. Using this form of notation will give the opportunity to 'play around' with the pretension and better understand what different pretensions would do to forces acting on the attachment designs and the participant.

```
F_FH_preTension = 14.0280  % Newton
F_FL_preTension = 129.2456 % Newton
F_BH_preTension = 0        % Newton
F_BL_preTension = 129.0894 % Newton
```

Figure 27: Preferred pretension (obtained in subchapter 4.1).

In Figure 28, the cable extension caused by the pretension is obtained using C= 160 N/m and the pretension mentioned in Figure 27. Also, the actual start lengths of the cables can now be computed in Figure 28 and are addressed as, e.g. lengthFH_preTension.

```
% as, F = C * u, where F is the pre-tension mentioned above and C = 160 N/m.
% The extention to reach this tension is u_FH_preTension = Force_FH_preTension / C
% so,
u_FH_preTension = F_FH_preTension / C
u_FL_preTension = F_FL_preTension / C
u_BH_preTension = F_BH_preTension / C
u_BL_preTension = F_BL_preTension / C

% so the start lengths of the cables (with pre-tension) need to be lengthCable_NOtension - u_Cable_preTension
lengthFH_preTension = lengthFH_NOtension - u_FH_preTension
lengthFL_preTension = lengthFL_NOtension - u_FL_preTension
lengthBH_preTension = lengthBH_NOtension - u_BH_preTension
lengthBL_preTension = lengthBL_NOtension - u_BL_preTension
```

Figure 28: extension by pretension and original lengths of the cables.

4.2 Extension (u) cables: lines 417-441

The cables' extensions during different gait cycle positions can be measured with the formulas in Figure 29. When the participant stands still in the middle of the treadmill, the extensions in the cables are the same as the pretension cable extension. However, in all other scenarios, the cable extension will be different. The second half of Figure 29 shows the

formulas for the cable extensions in the experimental setup. Again, either the top 4 formulas are used in the script or the bottom 4 formulas, depending on whether the extra items from the experimental set-up are incorporated. The ones that are not used are 'comment 'ed.

```
% Then the new u-cable should be (where the pre-tension is incorporated):
u_cableFH = (DistanceFH_RightAnkle + DistanceFH_LeftAnkle) - lengthFH_preTension
u_cableFL = (DistanceFLr_RightAnkle + DistanceFLl_LeftAnkle + CableLengthFL_middlepart) - lengthFL_preTension

u_cableBH = (DistanceBH_RightAnkle + DistanceBH_LeftAnkle) - lengthBH_preTension
u_cableBL = (DistanceBLr_RightAnkle + DistanceBLl_LeftAnkle + CableLengthBL_middlepart) - lengthBL_preTension

% --------------------------------
% In the experimental set-up with the hooks, carabiners, and force sensors:
 u_cableFH = (DistanceFH_RightAnkle + DistanceFH_LeftAnkle) - lengthFH_preTension - length_tensionhook
    - length_forcesensor - (3* length_carabiner)
 u_cableFL = (DistanceFLr_RightAnkle + DistanceFLl_LeftAnkle + CableLengthFL_middlepart) - lengthFL_preTension
    - length_tensionhook - length_forcesensor - (3* length_carabiner)

 u_cableBH = (DistanceBH_RightAnkle + DistanceBH_LeftAnkle) - lengthBH_preTension - length_tensionhook
    - length_forcesensor - (3* length_carabiner)
 u_cableBL = (DistanceBLr_RightAnkle + DistanceBLl_LeftAnkle + CableLengthBL_middlepart) - lengthBL_preTension
    - length_tensionhook - length_forcesensor - (3* length_carabiner)
```

Figure 29: Extension in the cables (pretension incorporated).

In all scenarios where a cable would slack, the script shows that the extension of that cable becomes negative. However, a negative extension means there is no tension in the cable. So, in Figure 30, the formulas are shown that change every negative extension to zero.

```
% When the extension (u) is a negative number, it means there is no tension in the cable.
% No tension means no force, so the following four formulas are included:
u_cableFH = max(u_cableFH,0)
u_cableFL = max(u_cableFL,0)
u_cableBH = max(u_cableBH,0)
u_cableBL = max(u_cableBL,0)
```

Figure 30: Negative extension becomes zero.

4.3 Forces (F) in cables

Forces in cables: lines 444-467

The tension in all four cables is computed with formula (7):

$$F = C * u \qquad (7)$$

The tension in the cable is equal to the pulling force it exerts on the right ankle and the pulling force on the left ankle. Figure 31 shows the above information but also adds the safety factor. To study the safety factor and its consequences, one only needs to change the number.

```
% GENERAL cable forces (in Newton)
Force_FH = C * u_cableFH      % Force in the Front high cable
Force_FL = C * u_cableFL      % Force in the Front low cable
Force_BH = C * u_cableBH      % Force in the Back high cable
Force_BL = C * u_cableBL      % Force in the Back low cable

% safety factor
Safety_factor = 1

% Cable forces pulling on right ankle (in Newton)
Force_FH_RightAnkle =  Force_FH * Safety_factor     % Force from the Front high cable on the right ankle
Force_FL_RightAnkle =  Force_FL * Safety_factor     % Force from the Front low cable on the right ankle
Force_BH_RightAnkle =  Force_BH * Safety_factor     % Force from the Back high cable on the right ankle
Force_BL_RightAnkle =  Force_BL * Safety_factor     % Force from the Back low cable on the right ankle

% Cable forces pulling on left ankle (in Newton)
Force_FH_LeftAnkle =  Force_FH * Safety_factor      % Force from the Front high cable on the left ankle
Force_FL_LeftAnkle =  Force_FL * Safety_factor      % Force from the Front low cable on the left ankle
Force_BH_LeftAnkle =  Force_BH * Safety_factor      % Force from the Back high cable on the left ankle
Force_BL_LeftAnkle =  Force_BL * Safety_factor      % Force from the Back low cable on the left ankle
```

Figure 31: forces in all four elastic cables.

Forces in cables, in x, y, and z-direction: lines 470-537

The tension in the cables mentioned in Figure 31 can be split into x, y, and z-directions. In the following Matlab calculations, the left and right ankles are considered separately because both ankles will be at different positions during the gait cycle.

In the formulas (8-10):
F = the pulling force in a cable on an ankle (see captions: “Cable forces pulling on right/left ankle (in Newton)”, in Figure 31).
Fx = the cable force pulling in the x-direction.
Fy = the cable force pulling in the y-direction.
Fz = the cable force pulling in the z-direction.
Alpha, beta, and gamma are the directional angles obtained with equations 4-6. All directional angles per cable and ankle can be found in Figure 16 and Appendix A.

$$F_x = F * \cos(alpha) \quad (8)$$

$$F_y = F * \cos(beta) \quad (9)$$

$$F_z = F * \cos(gamma) \quad (10)$$

Figure 32 shows the formulas for the pulling forces from the front high cable on the right ankle in the x, y, and z directions. The answers to these formulas are written down in vector form, addressed as the xyz force of the front high cable on the right ankle. All other xyz cable forces on the right ankle and on the left ankle can be found in Appendix B.

```
% Cable forces pulling on RIGHT ankle (in Newton)
% Front High Cable
Fx_R_FH = Force_FH_RightAnkle * cosd(alpha_R_FH)
Fy_R_FH = Force_FH_RightAnkle * cosd(beta_R_FH)
Fz_R_FH = Force_FH_RightAnkle * cosd(gamma_R_FH)

xyzForce_R_FH_check = [Fx_R_FH     Fy_R_FH     Fz_R_FH]
```

Figure 32: Front high cable force split into x, y, and z-direction.

Total pulling force on the attachments: lines 545-574

As mentioned at the beginning of the paper, there are eight cable ends divided into sets of two. The cable ends in the front on the right side are a set, and the ones in the front on the left are a set. The same holds for the cables in the back. Per set, the xyz cable forces can be added, so the total xyz pulling forces can be obtained for that specific set. Figure 33 shows the formulas, where each set represents either the front or the back of the left or right attachment.

```
% on the right attachement/ankle:
xyzForce_RightAnkle_front = xyzForce_R_FH_check + xyzForce_R_FLr_check   % Newton
xyzForce_RightAnkle_back  = xyzForce_R_BH_check + xyzForce_R_BLr_check  % Newton
% on the left attachement/ankle:
xyzForce_LeftAnkle_front = xyzForce_L_FH_check + xyzForce_L_FLl_check     % Newton
xyzForce_LeftAnkle_back  = xyzForce_L_BH_check + xyzForce_L_BLl_check     % Newton
```

Figure 33: xyz pulling forces per set of two cable ends.

Doing separate calculations for the left and right ankle/attachment is important, because of the differences in magnitude once the xyz forces are added. For example, the magnitude of the pulling force in the front high cable is the same on the left and right attachment, and the same is true for the front low cable. However, once the left xyz pulling forces of the front high and front low cable are added, and the right front high and low xyz pulling forces are added, these magnitudes start to differ. This is caused by the differences in Fx, Fy, and Fz at different ankle positions throughout the gait cycle. The magnitudes of the forces pulling on the front and back of the left and right attachment can be checked with the formulas from Figure 34.

```
% right attachment
totalForce_R_front = norm(xyzForce_RightAnkle_front)
totalForce_R_back = norm(xyzForce_RightAnkle_back)

% left attachment
totalForce_L_front = norm(xyzForce_LeftAnkle_front)
totalForce_L_back = norm(xyzForce_LeftAnkle_back)
```

Figure 34: Total pulling forces (magnitude) on the front/back of the right/left attachment.

Important note!

Before proceeding to the force distribution in the attachment designs, it is crucial to note that the original cable lengths from Kieft's paper, as depicted in Figure 19, were found to be too slack during use. Subsequent testing revealed that the cable lengths (in which the cables were found at the beginning of this current study) did not result in slack cables when walking with the MATE. Therefore, these lengths have replaced the values mentioned in Figure 19 in script 'nieuwWITHrighthandrule.m'. The new values of Figure 19 are:
OriginalCableLengthFH = 1.02 m
OriginalCableLengthFL = 1.62 m
OriginalCableLengthBH = 1.30 m
OriginalCableLengthBL = 2.02 m

These new original cable lengths affect the calculated pretension in the cables, obtained from the script 'nieuwWITHrighthandrule.m'. Therefore, the pretension values mentioned in Figure 22 are replaced with the following values.
Force_FH = 158.4 Newton
Force_FL = 163.98 Newton
Force_BH = 113.6 Newton
Force_BL = 99.83 Newton
Subsequently, these values are used as the pretension values in script `PreTensionCablesWITHrighthandrule.m' in lines 394 t/m 397, which means that Figure 27 changes to the values below.

```
F_FH_preTension = 158.3960  % Newton
F_FL_preTension = 163.9816 % Newton
F_BH_preTension = 113.5960 % Newton
F_BL_preTension = 99.8254 % Newton
```

5. Design force distribution

Choose left or right attachment: lines 576-614

As mentioned in the last chapter, the magnitude of the cable forces on the right and left cable end are the same. However, due to the different directions of the cable ends and the combination of the high and low cable pulling force on the attachment, the total pulling force (magnitude) on the left attachment can be different than the total pulling force (magnitude) on the right attachment. For this reason, the left and right attachments are analysed separately during the calculations of the force distribution on the attachment designs.

Figure 35 shows the pulling forces on the front and back of the right attachment in the x, y, and z directions. When analysing the left attachment, the script lines in Figure 35 need to be commented, and the lines in Figure 36 need to be uncommented.

Also, the forces in the front and back of the attachment have different magnitudes. Therefore, the pulling forces in the front and back are dealt with separately during the following calculations on force distribution.

```
% force distribution on the RIGHT attachement,
% use: xyzForce_RightAnkle_front & xyzForce_RightAnkle_back
F_max_front = xyzForce_RightAnkle_front
Fx_max_front = F_max_front(1,1)
Fy_max_front = F_max_front(1,2)
Fz_max_front = F_max_front(1,3)

% (use same attachement as above, add front or back), so if i alreally
% noted the RIGHT front force above, i now add the RIGHT back force as well.
F_max_back = xyzForce_RightAnkle_back
Fx_max_back = F_max_back(1,1)
Fy_max_back = F_max_back(1,2)
Fz_max_back = F_max_back(1,3)
```

Figure 35: pulling forces on the front/back of the right attachment.

```
% %If you want to know the force distribution on the LEFT attachemnt
% use: xyzForce_LeftAnkle_front & xyzForce_LeftAnkle_back)
% F_max_front = xyzForce_LeftAnkle_front
% Fx_max_front = F_max_front(1,1)
% Fy_max_front = F_max_front(1,2)
% Fz_max_front = F_max_front(1,3)
%
% F_max_back = xyzForce_LeftAnkle_back
% Fx_max_back = F_max_back(1,1)
% Fy_max_back = F_max_back(1,2)
% Fz_max_back = F_max_back(1,3)
```

Figure 36: pulling forces on the front/back of the left attachment.

Before the calculations on the force distribution can start, the angles in both designs need to be better analysed. Figure 37 shows the angle alpha in the cuff design, and Figures 38 and 39 show the angles in the bandage design. Figure 40 displays the angles in Matlab.



Figure 37: Cuff design.



Figure 38: Bandage design.

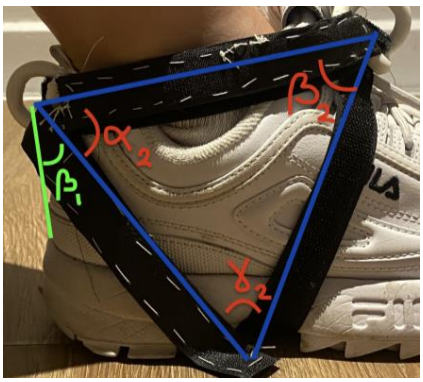


Figure 39: Bandage design.

```
% Angles in the designs (see fbd's), the same for left and right attachment.
Cuff_alpha = 30 % degrees

Bandage_alpha1 = 35   % degrees
Bandage_beta1 =  30   % degrees
% beta1 = 90 - alpha2
Bandage_alpha2 = 60   % degrees
Bandage_beta2 =  60   % degrees
% Bandage_gamma2 = 60 gedrees
```

Figure 40: Angles attachment designs (in Matlab).

5.1 Cuff design: lines 616-687

The cuff design, shown in Figure 41, consists of two components. The back part has a circular shape that wraps around the ankle like a cuff, and the front component is a small rectangular part that closes the circle from the front. The components are connected with the help of two Velcro straps located at the top of the cuff and at the bottom. The cuff design also contains two hooks that connect the design to the MATE cables. The hooks are placed in the middle of the front piece and on the back of the circular-shaped back piece. This subchapter elaborates on the force distribution across the cuff design. Calculations show how much force will be applied to the Velcro and how much surface area of Velcro is needed to withstand this force. Finally, the forces acting on the Velcro straps in worst-case scenarios for the cuff design are addressed.

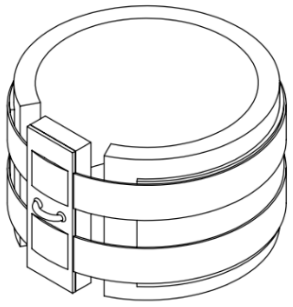


Figure 41: Cuff design.

Assumptions

- All the alpha angles are the same. See Figure 42.
- The xyz forces work directly on the design parts, not the hooks.
- The pulling force in the z-direction does not influence the Velcro straps in the xy plane.
- All forces are distributed equally over all the Velcro straps when pulling in the x-direction, so $F_{x,1} = F_{x,2}$. (See Figure 43.) In equations 11-13, these forces are referred to as F_x, so $F_{x,1} = F_{x,2} = F_x$. These are the x-components of the P shown in the free-body diagrams (FBDs).
- All forces are distributed equally over all the Velcro straps when pulling in the z-direction.
- The two big Velcro straps on the top and bottom of the cuff are divided into left and right parts, with the design component at the front as the dividing line. This results in four pieces of Velcro strap distributing the pulling forces across the cuff.
- P = pulling force in one strap.
- The cuff is in a static state in all the calculations in this subchapter.
- There is no friction between the cuff design and the participant wearing the cuff design.
- There is no rotation around the z-axis because the forces in the x-direction are always bigger than the forces in the y-direction.
- The hooks of the Velcro are equally resistant in each direction.

Worst case scenarios

- Having an open space between the circular piece and the front piece is less ideal, because in that case all pulling forces in y-direction are to be absorbed by the Velcro straps instead of partly being absorbed by the circular cuff component.

- The smaller the size of the rectangular front piece, the smaller the alpha, which increases the chance of the Velcro straps snapping, so a bigger front piece is preferred. However, having a small front piece and a bigger circular part ensures that there is enough space for the Velcro surface to distribute the pulling forces. So, a balance between these two options is required.
- As said, the two components in the cuff design are connected by Velcro straps. For this closing mechanism to work, at least either the top Velcro straps or the bottom Velcro straps must be securely closed. Otherwise, the cuff design will detach during use. So, one worst-case scenario would be when only two of the four Velcro strap pieces have to absorb all the pulling forces.

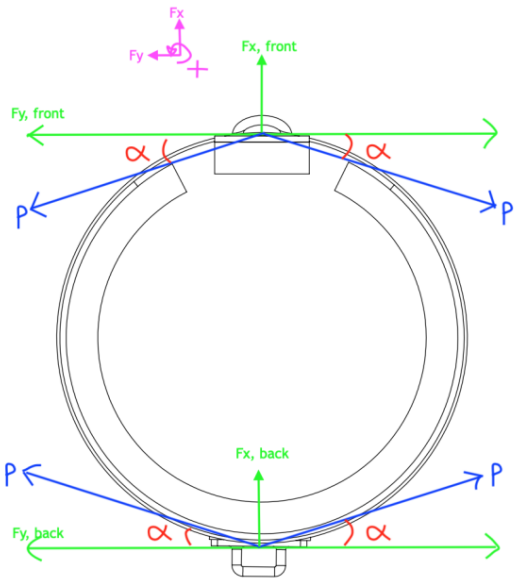


Figure 42: Cuff design in xy-plane

Force distribution x-direction

There are two forces in the x-direction pulling on the cuff: the pulling force on the front of the design and the one on the back. The biggest one of these two is used and is denoted by FxMAX. In the first rule of Figure 44, a vector with both the pulling forces in the x-direction, in the front and the back, is shown. In the next line, the biggest Fx of the two, in the absolute value, is selected and indicated by FxMAX. The absolute value is taken because the tension in the Velcro strap can't have a negative direction.

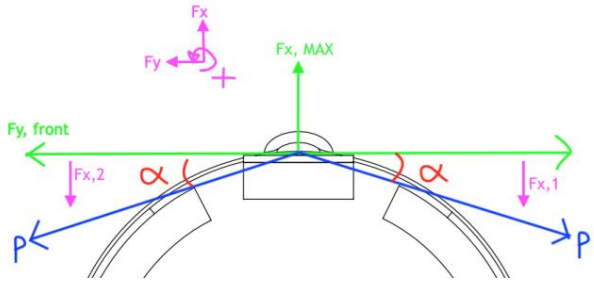


Figure 43: Force distribution Fx.

```
Fxfront_and_back = [Fx_max_front    Fx_max_back]
FxMAX = maxk(abs(Fxfront_and_back), 1)
```
Figure 44: Fx,max cuff design

The worst-case scenario for the Velcro straps from the x-direction is that the total pulling force needs to be distributed over two straps (as mentioned above). In that case, the force on each strap caused by Fx is calculated with equation 15. Figure 45 shows how this equation is mentioned in Matlab.

$$\sum F_x = 0 : F_{x,MAX} - F_{x,1} - F_{x,2} = 0 \quad (11)$$

$$F_{x,MAX} = 2F_x \quad (12)$$

$$F_x = \frac{1}{2} F_{x,MAX} \quad (13)$$

$$F_x = P * \sin(alpha) \quad (14)$$

$$P_{F_{x,worst-case}} = \frac{F_{x,MAX}}{2*\sin(alpha)} \quad (15)$$

```
WC_CuffStrap_by_x = FxMAX/(2*sind(Cuff_alpha))
```

Figure 45: force on each strap caused by Fx (worst-case scenario).

In a normal case scenario, the force in each strap caused by Fx,MAX is distributed across four straps. This results in equation 16 and the Matlab version shown in Figure 50.

$$P_{F_x} = \frac{F_{x,MAX}}{4*\sin(alpha)} \quad (16)$$

Force distribution y-direction

There are also two forces pulling on the cuff in the y-direction; see Figure 46. Both the Fy in the front and the Fy in the back of the cuff are used to determine the force P in the straps caused by Fy. However, in the case of Fy, either the two Velcro straps on the left of the front component or the two Velcro straps on the right are under tension. This results from the sum of the Fy,front and Fy,back pointing to either the left or the right side. When it points to the right, it causes slack in the right Velcro straps and tension in the left ones, etc. In the worst-case scenario, this means all of Fy must be covered by one strap instead of two (equation 20 and Figure 47).

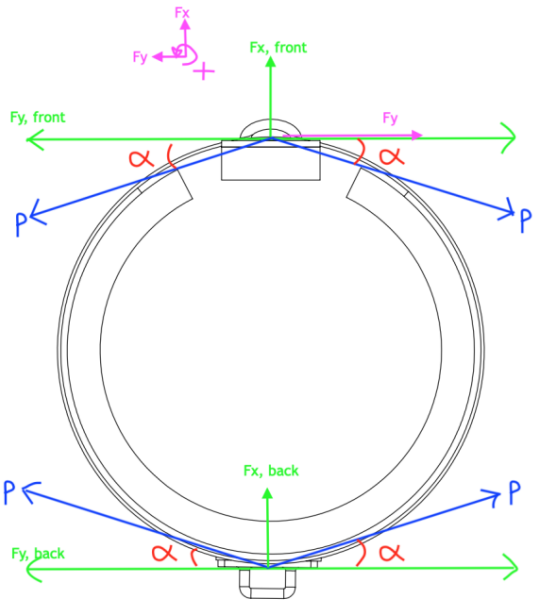


Figure 46: Force distribution Fy.

$$\sum F_y = 0: F_{y,front} + F_{y,back} - F_y = 0 \quad (17)$$

$$F_y = F_{y,front} + F_{y,back} \quad (18)$$

$$F_y = P * \cos(alpha) \quad (19)$$

$$P_{F_{y,worst-case}} = \frac{F_{y,front} + F_{y,back}}{\cos(alpha)} \quad (20)$$

```
WC_CuffStrap_by_y = abs(Fy_max_front + Fy_max_back)/cosd(Cuff_alpha)
```

Figure 47: force on left/right strap caused by Fy (worst-case scenario).

The absolute value is taken in the formula of Figure 47 because the tension in the straps cannot have a negative direction. In a normal case scenario, the force in the y-direction is absorbed by two Velcro straps instead of one. This results in the formula of equation 21 (shown in Figure 50).

$$P_{F_y} = \frac{F_{y,front} + F_{y,back}}{2*\cos(alpha)} \quad (21)$$

Force distribution z-direction

The pulling forces in the z-direction create a different kind of pulling force on the Velcro straps than the forces in the x and y directions. Therefore, they should be examined separately.

There are two forces in the z-direction, one in the front and one in the back of the cuff, see Figure 48. When the Fz,front and Fz, back would be in the same direction, the pulling forces on the Velcro straps would be minimal as both components would move in the same direction. This means the pulling force on the Velcro straps is biggest when Fz,front and Fz,back are in opposite directions. In the worst-case scenario, the pulling force is distributed over two straps instead of four (see equations 22-25 and Figure 49).

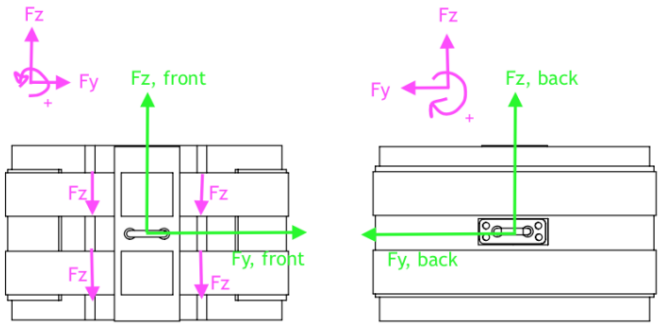


Figure 48: Force distribution Fz.

$$\sum F_z = 0: F_{z,front} - F_{z,back} - 2 * F_z = 0 \quad (22)$$

$$F_z = \frac{F_{z,front} - F_{z,back}}{2} \quad (23)$$

$$F_z = P \quad (24)$$

$$P_{F_{z,worst-case}} = \frac{F_{z,front} - F_{z,back}}{2} \quad (25)$$

```
WC_CuffStrap_by_z = abs(Fz_max_front - Fz_max_back)/2     % Newton.
```

Figure 49: force on each strap caused by Fz (worst-case scenario)

Again, the absolute value is taken in the formula (Figure 49) because the tension in the straps cannot have a negative direction. In the normal case scenario, the force in the z-direction is absorbed by all four straps, as shown in Figure 48. The formula matching this situation is shown in Equation 26 and Figure 50.

$$P_{F_z} = \frac{F_{z,front} - F_{z,back}}{4} \quad (26)$$

```
% In a normal scenario the pulling force x is distributed over 4 straps (instead of 2), so:
CuffStrap_by_x = WC_CuffStrap_by_x / 2
% In a normal scenario the pulling force y is distributed over 2 straps (instead of 1), so:
CuffStrap_by_y = WC_CuffStrap_by_y / 2
% In a normal scenario the pulling force z is distributed over 4 straps (instead of 2), so:
CuffStrap_by_z = WC_CuffStrap_by_z / 2
```

Figure 50: pulling force in straps (in normal scenarios).

Total tension in one strap caused by the pulling forces Px, Py, and Pz.

In the subchapter above, the pulling forces per strap, caused by the total pulling force in the x, y, and z-direction, are calculated separately. The total pulling force in a strap is a combination of these 'x, y, or z' - forces per strap. However, as mentioned above, the pulling forces in the z-direction create another kind of pulling force on the Velcro straps than the forces in the x and y directions. Therefore, they cannot be added and thus considered separately.

This results in two possible options for the total pulling force per strap. Option one: the total pulling force per strap is the sum of the pulling forces per strap from the x- and y-directions. Option two: the total pulling force per strap is equivalent to the pulling force per strap from the z-direction. The Velcro straps that hold the components of the attachment design together need to withstand at least the highest pulling force of the two options.

In a normal scenario, the two options are calculated as follows. The total pulling force in option one is the sum of equations 16 and 21. Even though not every strap experiences the same amount of pulling force because the pulling force from the y-direction only acts on the left or right cuff straps (see subchapter above), all four straps must be able to withstand this highest force. Therefore, the formula for every cuff strap is the sum of the pulling force from the x-direction and the pulling force from the y-direction (see equations 27-28).

$$Total\ force\ in\ one\ cuff\ strap = P_{F_x} + P_{F_y} \tag{27}$$

$$Total\ force\ in\ one\ cuff\ strap = \frac{F_{x,MAX}}{4 * \sin(alpha)} + \frac{F_{y,front} + F_{y,back}}{2 * \cos(alpha)} \tag{28}$$

The formula in a normal scenario for option two is equal to equation 26 and shown in equation 29.

$$Total\ force\ in\ one\ cuff\ strap = P_{F_z} = \frac{F_{z,front} - F_{z,back}}{4} \tag{29}$$

The Matlab function shown in Figure 51 is created to obtain the highest value of the two options easily.

```
% normal scenario of force distribution over the straps, then:
TotalForce_in_a_Cuffstrap = maxk([(CuffStrap_by_x + CuffStrap_by_y)     CuffStrap_by_z],1) % Newton.
```

Figure 51: total force in a cuff strap in a normal scenario.

In the worst-case scenario, the formula for the total pulling force on a strap is either equal to option one, shown in equation 31, or option two, visible in equation 32. Additionally, also for this scenario a Matlab function has been created to easily obtain the highest value of the two options, see figure 52.

$$Total\ force\ in\ one\ cuff\ strap_{worst-case} = P_{F_{x,worst-case}} + P_{F_{y,worst-case}} \tag{30}$$

$$Total\ force\ in\ one\ cuff\ strap_{worst-case} = \frac{F_{x,MAX}}{2 * \sin(alpha)} + \frac{F_{y,front} + F_{y,back}}{\cos(alpha)} \tag{31}$$

$$Total\ force\ in\ one\ cuff\ strap_{worst-case} = P_{F_{z,worst-case}} = \frac{F_{z,front} - F_{z,back}}{2} \quad (32)$$

```
WC_TotalForce_in_a_Cuffstrap = maxk([(WC_CuffStrap_by_x + WC_CuffStrap_by_y)      WC_CuffStrap_by_z],1)  % Newton.
% The total WC force in a strap is either the WC force in xy-direction or the WC force in z-direction.
```

Figure 52: total force in a cuff strap in the worst-case scenario.

Length Velcro strap

The amount of Velcro required in the straps of the attachment design to transfer the forces from the MATE to humans safely depends on its strength. The properties of the Velcro used for the cuff were not available. However, the properties of similar Velcro were found and used for the calculations.

```
% VELCRO STRAP PROPERTIES
ShearStrength_VelcroStrap_CuffDesign = 10.3;      % N/cm^2
FractureResistance_VelcroStrap_CuffDesign = 210; % N/cm
Width_Velcro_CuffDesign = 2;   % cm
```

The minimal Velcro strap length securing safe force transmission is calculated using the formula in Equation 33.

$$Length\ Velcro\ strap = \frac{Force}{width * shear\ strength} \quad (33)$$

```
% LENGTH VELCRO STRAP
% Length (in cm) = Force / (width * shear strength)
Length_of_one_cuffStrap = TotalForce_in_a_Cuffstrap /
(Width_Velcro_CuffDesign * ShearStrength_VelcroStrap_CuffDesign);
% Length of one Velcro strap in worst-case scenario
WC_Length_of_one_cuffStrap = WC_TotalForce_in_a_Cuffstrap /
(Width_Velcro_CuffDesign * ShearStrength_VelcroStrap_CuffDesign);
```

In the case of 'regular' walking (normal scenario 1, Figure 12), each Velcro strap should be at least 10.2 cm long. In the worst-case scenario (a strap detaching), each Velcro strap should be at least 20.4 cm long.

For user safety, it's crucial to understand the amount of force that could cause the Velcro straps on the cuff to break. The fracture resistance of a Velcro strap is calculated using Equation 34 to ensure that it exceeds the maximum force on one strap.

$$Fracture\ resistance\ Velcro\ strap\ cuff = fracture\ resistance\ Velcro * width\ Velcro \quad (34)$$

```
% FORCE OF FRACTURE VELCRO STRAP
% FractureResistance_VelcroStrap_CuffDesign = Force_FractureResistance /
Width_Velcro_CuffDesign
Force_FractureResistance_cuff = FractureResistance_VelcroStrap_CuffDesign *
Width_Velcro_CuffDesign;
```

The force at which a cuff Velcro strap breaks is 420 N. In the last line of the cuff design's Matlab script, this 420 N is compared with the total force on a cuff strap in the 'normal walk' scenario (normal scenario 1) and in the worst-case situation of a strap detaching in this scenario. In both conditions, the force on one strap is smaller than the breaking limit.

```
Compare_cuff = [Force_FractureResistance_cuff
TotalForce_in_a_Cuffstrap      WC_TotalForce_in_a_Cuffstrap];
```

5.2 Bandage design: lines 689-783

In this section, we analyse the force calculations for the bandage model. The bandage model, illustrated in Figure 53, includes an upper and lower strap. The upper strap wraps around the ankles, and the lower strap extends from the front of the foot, underneath it, to the back of the foot. Hooks are attached to the midpoints of the anterior and posterior of the ankle, connecting the cuff design to MATE's cables, and both the upper and lower straps are fastened to these hooks. The calculations presented in this subsection demonstrate the force applied to the Velcro straps of the bandage model and the necessary surface area of Velcro required to withstand this force. Furthermore, we address the forces acting on the Velcro straps in normal and worst-case scenarios for the bandage design.

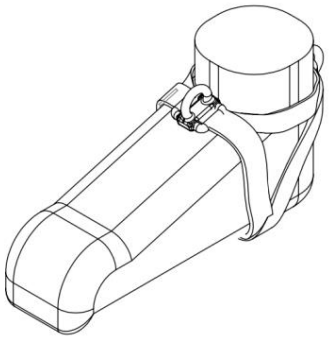


Figure 53: Bandage design.

Assumptions

- Assume that all points of the triangle in Figure 39 are fixed points.
- Assume that the angles alpha2, beta2, and gamma2 from Figure 39 add up to 180 degrees.
- Angles beta1 and alpha2 form a 90-degree angle.
- Assume equal force distribution in the bandage model on the right and left sides of the foot.
- The forces in the x-direction (frontal plane) and z-direction (transverse plane) acting on the front and back hooks of the design are equally split between both sides due to the 'equal distribution in the bandage model for both sides'.
- Assume that the Velcro is perfectly secure at both the upper and lower straps.
- Assume that the pulling forces act on specific points in the front and back of the design, directly on the straps and not via connecting hooks.
- Px = the pulling force acting on one side of the bandage model caused by the general pulling force in the x-direction (frontal plane). This Px value is used in the next paragraphs to calculate the force distribution in the lower and upper straps on that one side. In this context, Px,f represents the pulling force acting on the front part of the straps of one side of the bandage model and Px,b denotes the pulling force acting on the back part of the straps on that side of the design.
- Py = the pulling force acting on one side of the bandage model caused by the general pulling force in the y-direction (sagittal plane).

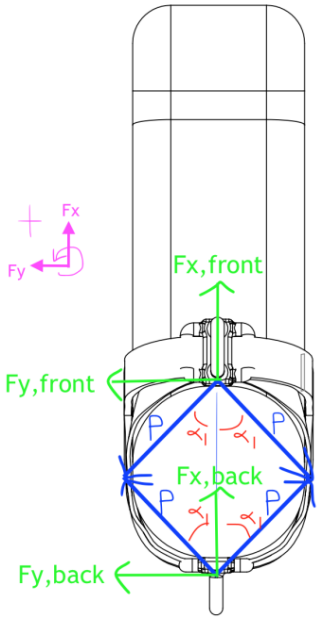


Figure 54: bandage model in xy-plane

- Pz = the pulling force acting on one side of the bandage model caused by the general pulling force in the z-direction (transverse plane).

Worst case scenarios

There are two worst-case scenarios associated with this model. The first scenario occurs when the lower strap fails, transferring all forces to the upper strap. The second scenario happens when the upper strap releases/fails, transferring all forces to the lower strap. The latter scenario is more likely to fail due to the construction of the lower strap wrapped around the foot, as opposed to the upper strap wrapped around the ankle.

Force distribution between the left and right sides of the bandage design

Prior to discussing the pulling forces exerted by the elastic cables of the MATE on the upper and lower straps of the bandage design, it is crucial to ascertain the magnitude of the force acting on each side of the design. Since the bandage wraps around the wearer's ankle and foot, it is essential to recognise that the left and right sides of the model do not lie on a flat plane. To initiate our analysis, we assume that the ankle, around which the bandage is wrapped, has a diamond shape (see Figure 54). This implies that the bandage design conforms to the diamond shape of the ankle. In our analysis, we assume that the forces on the right and left sides of the design are equal. We create separate free-body diagrams for each pulling force direction (x, y, and z) to examine how the force is distributed.

Force distribution in the x-direction

In the xy plane, the first free-body diagrams (FBDs) are examined. Two forces in the x-direction act on the attachment design: the pulling force on the front of the bandage model (Fx,f) and the force on the back (Fx,b). FBDs are provided for both the front and back of the design.

Figures 55 and 56 below illustrate that the anterior pulling force in the x-direction is evenly distributed over the left and right sides as Fx,x. Fx,x represents the x-component of Px,f, which allows us to calculate the value of Px,f.

The pulling force, caused by Fx,f, acting on the straps on one side of the model is Px,f.

$$\sum F_x = 0: F_{x,f} - F_{x,x} - F_{x,x} = 0 \quad (35)$$

$$\frac{F_{x,f}}{2} = F_{x,x} \quad (36)$$

$$F_{x,x} = P_{x,f} * \cos(alpha1) \quad (37)$$

$$P_{x,f} = \frac{F_{x,f}}{2 * \cos(alpha1)} \quad (38)$$

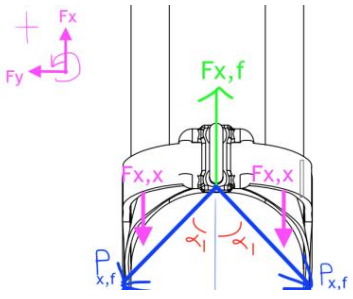


Figure 55: Force distribution Fx,f.

And, the pulling force, caused by Fx,b, acting on the straps on that one side of the model is Px,b.

$$\sum F_x = 0: -F_{x,b} + F_{x,x} + F_{x,x} = 0 \quad (39)$$

$$\frac{F_{x,b}}{2} = F_{x,x} \quad (40)$$

$$F_{x,x} = P_{x,b} * \cos(alpha1) \quad (41)$$

$$P_{x,b} = \frac{F_{x,b}}{2 * cos(alpha1)} \quad (42)$$

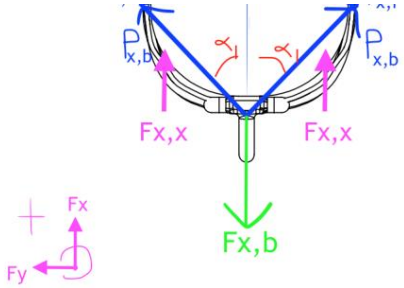


Figure 56: Force distribution Fx,b.

Force distribution in y-direction

In the same xy-plane, the FBDs of the MATE's pulling forces in the y-direction are shown on the front and back of the design (Fy,f and Fy,b). In Figures 57 and 58, these FBDs are demonstrated. Where Fy,y indicates the y-component of Py,f in Figure 57, and the y-component of Py,b in Figure 58.

The Fy,f affects only one side of the bandage model as one side receives all the pulling force (Fy,y) depending on the direction of the y-force. Only taking into account this y-force means that one side of the straps is under tension while the other side is slack. How this pulling force Fy,f is transferred to one side of the bandage design is indicated by Py,f.

$$\sum F_y = 0: F_{y,f} - F_{y,y} = 0 \quad (43)$$

$$F_{y,f} = F_{y,y} \quad (44)$$

$$F_{y,y} = P_{y,f} * \sin(alpha1) \quad (45)$$

$$P_{y,f} = \frac{F_{y,f}}{\sin(alpha1)} \quad (46)$$

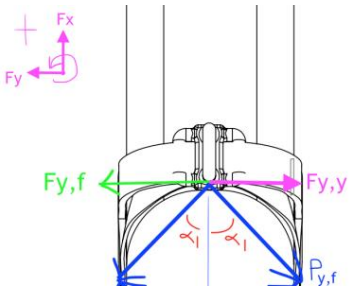


Figure 57: Force distribution Fy,f.

Py, b indicates how the pulling force Fy,b is transferred to the bandage design.

$$\sum F_y = 0: F_{y,b} - F_{y,y} = 0 \quad (47)$$

$$F_{y,b} = F_{y,y} \quad (48)$$

$$F_{y,y} = P_{y,b} * \sin(alpha1) \quad (49)$$

$$P_{y,b} = \frac{F_{y,b}}{\sin(alpha1)} \quad (50)$$

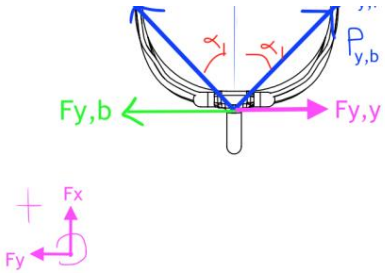


Figure 58: Force distribution Fy,b.

Force distribution in z-direction

The FBDs of the pulling forces on the design in z-direction are presented in the yz plane. The force Fz,f on the anterior of the foot is shown in Figure 59, and the pulling force Fz,b on the posterior of the foot is illustrated in Figure 60. With the help of their z-components (Fz,z) Pz,f and Pz,b, respectively, show how the Fz forces are distributed over the sides of the bandage design.

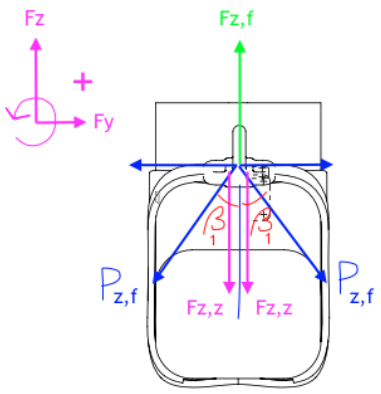


Figure 59: Force distribution Fz,f.

$$\sum F_z = 0: F_{z,f} - F_{z,z} - F_{z,z} = 0 \quad (51)$$

$$\frac{F_{z,f}}{2} = F_{z,z} \quad (52)$$

$$F_{z,z} = P_{z,f} * \cos(beta1) \quad (53)$$

$$P_{z,f} = \frac{F_{z,f}}{2*\cos(beta1)} \quad (54)$$

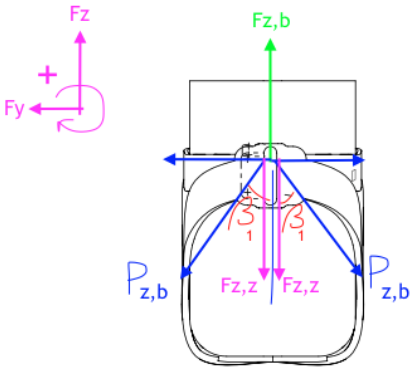


Figure 60: Force distribution Fz,b.

$$\sum F_z = 0: F_{z,b} - F_{z,z} - F_{z,z} = 0 \quad (55)$$

$$\frac{F_{z,b}}{2} = F_{z,z} \quad (56)$$

$$F_{z,z} = P_{z,b} * \cos(beta1) \quad (57)$$

$$P_{z,b} = \frac{F_{z,b}}{2*\cos(beta1)} \quad (58)$$

The Matlab calculating the equations 35-58 are shown below.

```
% xy-FBD:
% for forces in x-direction this means:
Px_front = Fx_max_front / (2 * cosd(Bandage_alpha1))
Px_back  = abs(Fx_max_back) / (2 * cosd(Bandage_alpha1)) % put abs around
Fx_max_back because the pulling force in the design is not negative, only
the direction of the cable force is.

% for forces in y-direction this means:
Py_front = Fy_max_front / sind(Bandage_alpha1) % either direction of
Fy_max_front will cause a positive pulling force in one side of the design,
so use abs(Py_front) in formula below.
Py_back  = Fy_max_back / sind(Bandage_alpha1)  % either direction of
Fy_max_back will cause a positive pulling force in one side of the design,
so use abs(Py_back) in formula below.

% yz-FBD:
Pz_front = Fz_max_front / (2 * cosd(Bandage_beta1))
Pz_back = Fz_max_back / (2 * cosd(Bandage_beta1))
```

Total tension in one strap caused by the pulling forces

Only taking into account the forces in x and z direction.

Now that we have established the distribution of MATE's force on both sides of the design, we can investigate how the force is distributed between the upper and lower straps. To do this, we analyse the free-body diagram of the right side of the foot in the xz plane. The forces determined from the previous paragraph are the forces exerted by MATE on this side of the design, so these forces are utilised in subsequent free-body diagrams to represent the forces acting in the x, y, and z directions. A force overview is shown in Figure 61.

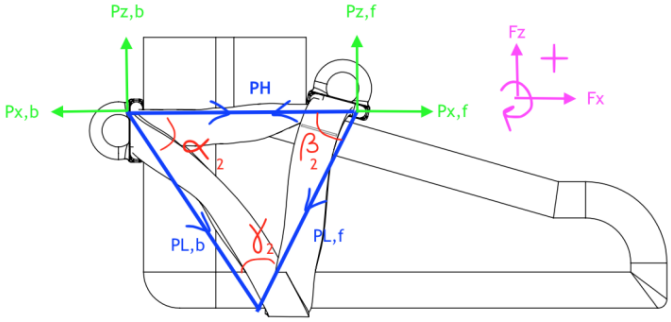
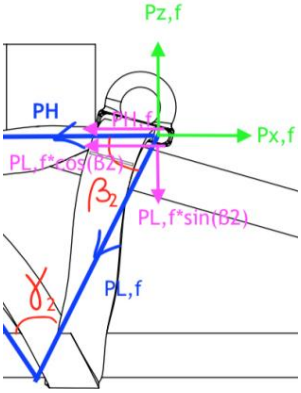


Figure 61: Force distribution in the xz plane on the right side of the bandage model.

In Figure 62, the FBD of the anterior part of the design is analysed. PL,f denotes the pulling force in the lower strap, while PH,f signifies the forces in the upper strap resulting from the pulling forces in the x and z directions (Pz,f and Px,f). In Figure 63, a similar FBD is presented for the posterior part of the design, enabling the determination of the pulling forces on the upper and lower straps, PH,b and PL,b.



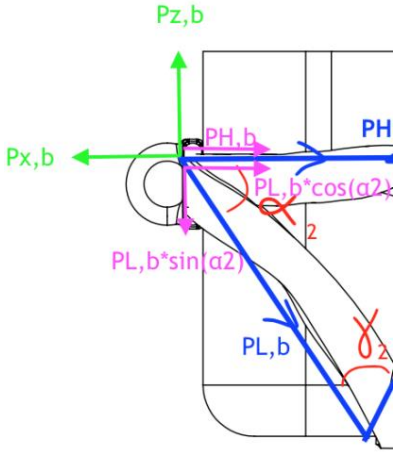
$$\sum F_z = 0: P_{z,f} - P_{L,f} * \sin(beta2) = 0 \quad (59)$$

$$P_{L,f} = \frac{P_{z,f}}{\sin(beta2)} \quad (60)$$

$$\sum F_x = 0: P_{x,f} - P_{H,f} - P_{L,f} * \cos(beta2) = 0 \quad (61)$$

$$P_{H,f} = P_{x,f} - \frac{P_{z,f} * \cos(beta2)}{\sin(beta2)} \quad (62)$$

Figure 62: Obtaining PH,f and PL,f.



$$\sum F_z = 0: P_{z,b} - P_{L,b} * \sin(alpha2) = 0 \quad (63)$$

$$P_{L,b} = \frac{P_{z,b}}{\sin(alpha2)} \quad (64)$$

$$\sum F_x = 0: P_{H,b} + P_{L,b} * \cos(alpha2) - P_{x,b} = 0 \quad (65)$$

$$P_{H,b} = P_{x,b} - \frac{P_{z,b} * \cos(alpha2)}{\sin(alpha2)} \quad (66)$$

Figure 63: Obtaining PH,b and PL,b

Adding the pulling forces of the y-direction.

Given that the points of the triangle in Figure 61 are fixed (an assumption) and there is no exchange of pulling forces between both sides of the bandage model, it follows that the pulling forces on both sides of the bandage model may not be equal due to the Py pulling forces working on the design. However, since Py can change direction, all straps must be able to endure the highest pulling force exerted on them.

The Py forces are assumed to be evenly distributed over the upper and lower straps. Consequently, 0.5Py is included in the pulling force per strap equations of 60, 62, 64, and 66.

$$P_{L,f_with_Py} = \frac{P_{z,f}}{\sin(beta2)} + \left|\frac{P_{y,f}}{2}\right| \quad (67)$$

$$P_{H,f_with_Py} = P_{x,f} - \frac{P_{z,f}*\cos(beta2)}{\sin(beta2)} + \left|\frac{P_{y,f}}{2}\right| \quad (68)$$

$$P_{L,b_with_Py} = \frac{P_{z,b}}{\sin(alpha2)} + \left|\frac{P_{y,b}}{2}\right| \quad (69)$$

$$P_{H,b_with_Py} = P_{x,b} - \frac{P_{z,b}*\cos(alpha2)}{\sin(alpha2)} + \left|\frac{P_{y,b}}{2}\right| \quad (70)$$

The corresponding lines in Matlab are shown below.

```
% Assume both sides of the attachments are equal in design, in the sagittal
plane (split on the x-axis).
% xz-FBD, without Py (in Newton):
PL_front =  Pz_front / sind(Bandage_beta2)
PL_front = max(PL_front,0) % because when PL_front becomes negative, there
is no tension anymore in PL_front cable. So, the pulling force in the cable
PL_front is set to 0.
PH_front = Px_front - (PL_front * cosd(Bandage_beta2))

PL_back = Pz_back / sind(Bandage_alpha2)
PL_back = max(PL_back,0) % because when PL_back becomes negative, there is
no tension anymore in PL_back cable. So, the pulling force in the cable
PL_back is set to 0.
PH_back = Px_back - (PL_back * cosd(Bandage_alpha2))

% include Py
PL_front_WITHPy =  (Pz_front / sind(Bandage_beta2)) + abs(0.5*Py_front)
PL_front_WITHPy = max(PL_front_WITHPy,0) % because when PL_front becomes
negative, there is no tension anymore in PL_front cable. So, the pulling
force in the cable PL_front is set to 0.
PH_front_WITHPy = (Px_front - (PL_front_WITHPy * cosd(Bandage_beta2))) +
abs(0.5*Py_front)

PL_back_WITHPy = (Pz_back / sind(Bandage_alpha2)) + abs(0.5*Py_back)
PL_back_WITHPy = max(PL_back_WITHPy,0) % because when PL_back becomes
negative, there is no tension anymore in PL_back cable. So, the pulling
force in the cable PL_back is set to 0.
PH_back_WITHPy = (Px_back - (PL_back_WITHPy * cosd(Bandage_alpha2))) +
abs(0.5*Py_back)
```

Pulling force in the upper strap and lower strap

In the course of the force calculations, a distinction was made between the anterior and posterior of the bandage model. Nevertheless, the pulling forces obtained in both scenarios act on the same straps. (PH,b and PH,f pull on the same strap, and so do PL,b and PL,f.) Consequently, the higher of the two forces is considered as the value of the pulling force in that particular strap. Here, PL represents the pulling force in the lower strap, and PH represents the pulling force in the upper strap.

```
% the highest PL and PH are (in Newton):
PH_max = maxk([PH_back_WITHPy    PH_front_WITHPy],1)
PL_max = maxk([PL_back_WITHPy    PL_front_WITHPy],1)
```

Pulling forces in the upper and lower strap in the worst-case scenarios

As mentioned earlier, the worst-case scenarios occur when all the pulling force is concentrated on either the upper straps or the lower straps. We are still examining only one side of the bandage model.

Worst-case scenario 1: all pulling force on the upper strap

In this worst-case scenario, only the sum of the forces in the x and y directions is taken into account, with the pulling force in the z-direction and the upper strap being perpendicular to each other. Once again, the highest value of (Px,b + Py,b) and (Px,f + Py,f) is chosen.

$$P_{UpperStrap\,worst-case} = P_{x,b} + P_{y,b}\ OR = P_{x,f} + P_{y,f} \qquad (71)$$

```
% Worst case scenario 1: all pulling force on the upper straps.
WC_upperStrap = maxk([(Px_back + abs(Py_back))    (Px_front +
abs(Py_front))],1)    % Newton, force on the one side of the upper strap)
```

Worst-case scenario 2: all pulling force on the lower strap

The scenario in which the upper strap detaches and all the force pulls on the lower strap, is more prone to failure than the reverse situation. This is because a secure upper strap is crucial for maintaining the integrity of the wrapped construction of the lower strap.

If Pz>0, then:

$P_{x,f} = P_{L,f} * \cos(beta2)$ and Py,f is added to PL,f, so: $P_{L,f} = \frac{P_{x,f}}{\cos(beta2)} + |P_{y,f}|$ (72)

$P_{z,f} = P_{L,f} * \sin(beta2)$ and Py,f is added to PL,f, so: $P_{L,f} = \frac{P_{z,f}}{\sin(beta2)} + |P_{y,f}|$ (73)

$P_{x,b} = P_{L,b} * \cos(alpha2)$ and Py,b is added to PL,b, so: $P_{L,b} = \frac{P_{x,b}}{\cos(alpha2)} + |P_{y,b}|$ (74)

$P_{z,b} = P_{L,b} * \sin(alpha2)$ and Py,b is added to PL,b, so: $P_{L,b} = \frac{P_{z,b}}{\sin(alpha2)} + |P_{y,b}|$ (75)

The highest value of the four equations mentioned above is the worst-case pulling force on the lower strap ($P_{LowerStrap\,worst-case}$) when the upper strap is detached.

If Pz is less than 0, it is highly likely that the bandage design will detach completely from the foot. In cases where the negative pulling force is very low, there's a chance that friction with the shoe and the higher pulling force in the x-direction will delay the detachment process, but ultimately the design will detach.

```
% Worst case scenario 2: all pulling force on the lower straps.
Max_xzforce_in_lowerStrap_front = maxk([(Px_front/cosd(Bandage_beta2))
(Pz_front/sind(Bandage_beta2))],1)
Max_xzforce_in_lowerStrap_back = maxk([(Px_back/cosd(Bandage_alpha2))
(Pz_back/sind(Bandage_alpha2))],1)
WC_lowerStrap = maxk([(Max_xzforce_in_lowerStrap_back + abs(Py_back))
(Max_xzforce_in_lowerStrap_front + abs(Py_front))],1)    % Newton, force on
the one side of the lower strap.
```

Length Velcro strap

As stated previously, the quantity of Velcro required in the attachment design's straps to safely transfer forces from the MATE to humans depends on its strength. Since the properties of the Velcro used in the bandage design were not available, we utilised the properties of similar Velcro for the calculations.

```
% VELCRO STRAP PROPERTIES
ShearStrength_VelcroStrap_BandageDesign = 7.0    % N/cm^2
FractureResistance_VelcroStrap_BandageDesign = 120 % N/cm^2
Width_Velcro_BandageDesign = 2  % cm
Thickness_Veclro_BandageDesign = 0.135 % cm
```

The minimal Velcro strap length securing safe force transmission is calculated using the formulas in Equations 76 and 77 .

$$Length\ Velcro\ upper\ strap = \frac{PH_{Max}}{width * shear\ strength} \tag{76}$$

$$Length\ Velcro\ lower\ strap = \frac{PL_{Max}}{width * shear\ strength} \tag{77}$$

In the case of 'regular' walking (normal scenario 1, Figure 12), each upper Velcro strap should be a minimum of 14.3 cm long, and each lower Velcro strap should be at least 9.6 cm long. In the worst-case scenario where the lower strap detaches, each Velcro upper strap should be a minimum of 21.7 cm long. In the worst-case scenario of the upper strap detaching, there should be at least 33.7 cm of Velcro per lower strap to withstand the pulling forces.

```
% LENGTH VELCRO STRAP
% Length (in cm) = Force / (width * shear strength)
Length_of_one_upperBANDAGEStrap = PH_max / (Width_Velcro_BandageDesign *
ShearStrength_VelcroStrap_BandageDesign) % cm
Length_of_one_lowerBANDAGEStrap = PL_max / (Width_Velcro_BandageDesign *
ShearStrength_VelcroStrap_BandageDesign) % cm

% Length of one Velcro strap in worst case scenario (in cm)
WC1_Length_of_one_upperBANDAGEStrap = WC_upperStrap /
(Width_Velcro_BandageDesign * ShearStrength_VelcroStrap_BandageDesign) % cm
WC2_Length_of_one_lowerBANDAGEStrap = WC_lowerStrap /
(Width_Velcro_BandageDesign * ShearStrength_VelcroStrap_BandageDesign) % cm
```

Also in the bandage design, we verify the amount of force necessary to break the Velcro straps to ensure user safety. The fracture resistance of the Velcro strap is calculated using the equation below to guarantee that it can withstand the maximum forces applied to the bandage design.

$$Fracture\ resistance\ Velcro\ strap\ bandage = fracture\ resistance\ Velcro * width\ Velcro * 1\ cm \quad (78)$$

The Velcro strap breaks at a pulling force of 240 Newtons. Under normal walking conditions, the pulling forces remain below this limit. However, in worst-case scenarios, the force can exceed this threshold. In the bandage design, the Velcro is attached to lashing straps that can withstand pulling forces of over 1200 Newtons, ensuring that strap fracture is not a concern.

```
% FORCE OF FRACTURE VELCRO STRAP
% Checking if our fracture resistance force is still higher than max force
on one strap.
Force_FractureResistance_bandage =
FractureResistance_VelcroStrap_BandageDesign * Width_Velcro_BandageDesign

% Compare the 'Force_FractureResistance' one strap can handle to the 'total
force in a bandage strap', and the 'worst case scenario total force in a
bandage strap':
Compare_bandageDesign = [Force_FractureResistance_bandage       PH_max
WC_upperStrap;
                         Force_FractureResistance_bandage       PL_max
WC_lowerStrap]
```

Appendices

Appendix A

Figure AA elaborates on how Equations 4-6 are transferred to Matlab.

```
% Below an explanation of the used formula's:
% v = .... direction vector (cable), see: RIGHT/LEFT ankle to frame (directional vector: in x, y and z direction)
% i = [1 0 0] % unit vector for positive x-axis
% j = [0 1 0] % unit vector for positive y-axis
% k = [0 0 1] % unit vector for positive z-axis
%
% % direction angles
% alpha = acosd(dot(v,i)/(norm(v)*norm(i))) % angle formed between vector and positive x-axis, in degrees
% beta = acosd(dot(v,j)/(norm(v)*norm(j))) % angle formed between vector and positive y-axis, in degrees
% gamma = acosd(dot(v,k)/(norm(v)*norm(k))) % angle formed between vector and positive z-axis, in degrees
%
% angles_degrees = [alpha beta gamma]
```

Figure AA: example formulas for calculating the directional angles in Matlab.

Figure AB shows all directional angles (in degrees) from the right ankle to the frame. The corresponding directional vectors are mentioned in Figure 15.

```
% Right ankle to frame:
% Front High
alpha_R_FH = acosd(dot(RightAnkle_to_FrameFH,i)/(norm(RightAnkle_to_FrameFH)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_R_FH = acosd(dot(RightAnkle_to_FrameFH,j)/(norm(RightAnkle_to_FrameFH)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_R_FH = acosd(dot(RightAnkle_to_FrameFH,k)/(norm(RightAnkle_to_FrameFH)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_R_FH_degrees = [alpha_R_FH   beta_R_FH   gamma_R_FH]

% Front Low (right)
alpha_R_FLr = acosd(dot(RightAnkle_to_FrameFLr,i)/(norm(RightAnkle_to_FrameFLr)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_R_FLr = acosd(dot(RightAnkle_to_FrameFLr,j)/(norm(RightAnkle_to_FrameFLr)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_R_FLr = acosd(dot(RightAnkle_to_FrameFLr,k)/(norm(RightAnkle_to_FrameFLr)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_R_FLr_degrees = [alpha_R_FLr     beta_R_FLr  gamma_R_FLr]

% Back High
alpha_R_BH = acosd(dot(RightAnkle_to_FrameBH,i)/(norm(RightAnkle_to_FrameBH)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_R_BH = acosd(dot(RightAnkle_to_FrameBH,j)/(norm(RightAnkle_to_FrameBH)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_R_BH = acosd(dot(RightAnkle_to_FrameBH,k)/(norm(RightAnkle_to_FrameBH)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_R_BH_degrees = [alpha_R_BH    beta_R_BH   gamma_R_BH]

% Back Low (right)
alpha_R_BLr = acosd(dot(RightAnkle_to_FrameBLr,i)/(norm(RightAnkle_to_FrameBLr)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_R_BLr = acosd(dot(RightAnkle_to_FrameBLr,j)/(norm(RightAnkle_to_FrameBLr)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_R_BLr = acosd(dot(RightAnkle_to_FrameBLr,k)/(norm(RightAnkle_to_FrameBLr)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_R_BLr_degrees = [alpha_R_BLr   beta_R_BLr  gamma_R_BLr]
```

Figure AB: the directional angles of all the directional vectors from the right ankle to the pulleys on the frame in Matlab.

In Figure AC, all formulas for the directional vectors, directional angles, and absolute distances of the LEFT ankle to the corresponding pulleys on the frame can be found.

```
% ------------------------------------------------------------------------
% LEFT ankle to frame (directional vector: in x, y and z direction)
LeftAnkle_to_FrameFH = [FH - AnkleL]      % meters
LeftAnkle_to_FrameFLl = [FLl - AnkleL]    % meters

LeftAnkle_to_FrameBH = [BH - AnkleL]      % meters
LeftAnkle_to_FrameBLl = [BLl - AnkleL]    % meters

% LEFT ankle to frame: DIRECTIONAL ANGLES--------------------------------
% Left ankle to frame: Front High
alpha_L_FH = acosd(dot(LeftAnkle_to_FrameFH,i)/(norm(LeftAnkle_to_FrameFH)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_L_FH = acosd(dot(LeftAnkle_to_FrameFH,j)/(norm(LeftAnkle_to_FrameFH)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_L_FH = acosd(dot(LeftAnkle_to_FrameFH,k)/(norm(LeftAnkle_to_FrameFH)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_L_FH_degrees = [alpha_L_FH   beta_L_FH   gamma_L_FH]

% Front Low (left)
alpha_L_FLl = acosd(dot(LeftAnkle_to_FrameFLl,i)/(norm(LeftAnkle_to_FrameFLl)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_L_FLl = acosd(dot(LeftAnkle_to_FrameFLl,j)/(norm(LeftAnkle_to_FrameFLl)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_L_FLl = acosd(dot(LeftAnkle_to_FrameFLl,k)/(norm(LeftAnkle_to_FrameFLl)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_L_FLl_degrees = [alpha_L_FLl     beta_L_FLl    gamma_L_FLl]

% Back High
alpha_L_BH = acosd(dot(LeftAnkle_to_FrameBH,i)/(norm(LeftAnkle_to_FrameBH)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_L_BH = acosd(dot(LeftAnkle_to_FrameBH,j)/(norm(LeftAnkle_to_FrameBH)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_L_BH = acosd(dot(LeftAnkle_to_FrameBH,k)/(norm(LeftAnkle_to_FrameBH)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_L_BH_degrees = [alpha_L_BH     beta_L_BH    gamma_L_BH]

% Back Low (left)
alpha_L_BLl = acosd(dot(LeftAnkle_to_FrameBLl,i)/(norm(LeftAnkle_to_FrameBLl)*norm(i))) % angle formed between vector and positive x-axis, in degrees
beta_L_BLl = acosd(dot(LeftAnkle_to_FrameBLl,j)/(norm(LeftAnkle_to_FrameBLl)*norm(j))) % angle formed between vector and positive y-axis, in degrees
gamma_L_BLl = acosd(dot(LeftAnkle_to_FrameBLl,k)/(norm(LeftAnkle_to_FrameBLl)*norm(k))) % angle formed between vector and positive z-axis, in degrees

angles_L_BLl_degrees = [alpha_L_BLl     beta_L_BLl    gamma_L_BLl]

% LEFT ankle to frame (absolute distance)----------------------------------
DistanceFH_LeftAnkle = norm([FH - AnkleL])     % meters
DistanceFLl_LeftAnkle = norm([FLl - AnkleL])   % meters

DistanceBH_LeftAnkle = norm([BH - AnkleL])     % meters
DistanceBLl_LeftAnkle = norm([BLl - AnkleL])   % meters
```

Figure AC: the directional vectors, directional angles, and the absolute distance of the LEFT ankle to the corresponding pulleys on the frame in Matlab.

Appendix B

Figure BA shows all cable forces on the right and left ankle split into x, y, and z directions.

```
% CABLE FORCES DISTRIBUTED IN X, Y, AND Z-DIRECTION

% Below: an explanation of the used formula's!!
% F = ... % pulling force in cable (on an ankle), see: Cable forces pulling on right/left ankle (in Newton)
% Fx = F * cosd(alpha)
% Fy = F * cosd(beta)
% Fz = F * cosd(gamma)
%
% F_check = [Fx Fy Fz]

% Cable forces pulling on RIGHT ankle (in Newton)
% Front High Cable
Fx_R_FH = Force_FH_RightAnkle * cosd(alpha_R_FH)
Fy_R_FH = Force_FH_RightAnkle * cosd(beta_R_FH)
Fz_R_FH = Force_FH_RightAnkle * cosd(gamma_R_FH)

xyzForce_R_FH_check = [Fx_R_FH     Fy_R_FH     Fz_R_FH]

% Front Low right Cable
Fx_R_FLr = Force_FL_RightAnkle * cosd(alpha_R_FLr)
Fy_R_FLr = Force_FL_RightAnkle * cosd(beta_R_FLr)
Fz_R_FLr = Force_FL_RightAnkle * cosd(gamma_R_FLr)

xyzForce_R_FLr_check = [Fx_R_FLr   Fy_R_FLr  Fz_R_FLr]

% Back High Cable
Fx_R_BH = Force_BH_RightAnkle * cosd(alpha_R_BH)
Fy_R_BH = Force_BH_RightAnkle * cosd(beta_R_BH)
Fz_R_BH = Force_BH_RightAnkle * cosd(gamma_R_BH)

xyzForce_R_BH_check = [Fx_R_BH   Fy_R_BH  Fz_R_BH]

% Back Low right Cable
Fx_R_BLr = Force_BL_RightAnkle * cosd(alpha_R_BLr)
Fy_R_BLr = Force_BL_RightAnkle * cosd(beta_R_BLr)
Fz_R_BLr = Force_BL_RightAnkle * cosd(gamma_R_BLr)

xyzForce_R_BLr_check = [Fx_R_BLr   Fy_R_BLr  Fz_R_BLr]

% Cable forces pulling on LEFT ankle (in Newton)
% Front High Cable
Fx_L_FH = Force_FH_LeftAnkle * cosd(alpha_L_FH)
Fy_L_FH = Force_FH_LeftAnkle * cosd(beta_L_FH)
Fz_L_FH = Force_FH_LeftAnkle * cosd(gamma_L_FH)

xyzForce_L_FH_check = [Fx_L_FH    Fy_L_FH  Fz_L_FH]

% Front Low left Cable
Fx_L_FLl = Force_FL_LeftAnkle * cosd(alpha_L_FLl)
Fy_L_FLl = Force_FL_LeftAnkle * cosd(beta_L_FLl)
Fz_L_FLl = Force_FL_LeftAnkle * cosd(gamma_L_FLl)

xyzForce_L_FLl_check = [Fx_L_FLl    Fy_L_FLl  Fz_L_FLl]

% Back High Cable
Fx_L_BH = Force_BH_LeftAnkle * cosd(alpha_L_BH)
Fy_L_BH = Force_BH_LeftAnkle * cosd(beta_L_BH)
Fz_L_BH = Force_BH_LeftAnkle * cosd(gamma_L_BH)

xyzForce_L_BH_check = [Fx_L_BH   Fy_L_BH  Fz_L_BH]

% Back Low left Cable
Fx_L_BLl = Force_BL_LeftAnkle * cosd(alpha_L_BLl)
Fy_L_BLl = Force_BL_LeftAnkle * cosd(beta_L_BLl)
Fz_L_BLl = Force_BL_LeftAnkle * cosd(gamma_L_BLl)

xyzForce_L_BLl_check = [Fx_L_BLl    Fy_L_BLl  Fz_L_BLl]
```

Figure BA: cable forces split into x, y, and z-directions.

References force calculations

1. L. Kieft, “Design validation of the mate tendon-based passive device for post-stroke rehabilitation,” Oct 2022
2. P. Rijndam, “Stroke rehabilitation on the lower body,” Jun 2023
3. Balaban, B., & Tok, F. (2014). Gait disturbances in patients with stroke. *PM & R: the journal of injury, function, and rehabilitation*, *6*(7), 635–642. https://doi.org/10.1016/j.pmrj.2013.12.017
4. Jurca, A., Žabkar, J., & Džeroski, S. (2019). Analysis of 1.2 million foot scans from North America, Europe and Asia. *Scientific Reports*, *9*(1). https://doi.org/10.1038/s41598-019-55432-z
5. Jung, W. C., & Lee, J. K. (2021). Treadmill-to-Overground Mapping of Marker Trajectory for Treadmill-Based Continuous Gait Analysis. *Sensors*, *21*(3), 786. https://doi.org/10.3390/s21030786
6. Whittle, M. W. (2007). *Stride Width - an overview | ScienceDirect Topics*. Www.sciencedirect.com. https://www.sciencedirect.com/topics/nursing-and-health-professions/stride-width#:~:text=Normal%20stride%20width%20for%20adults
7. Herrero-Larrea, A., Miñarro, A., Narvaiza, L., Gálvez-Barrón, C., León, N. G., Valldosera, E., Felipe, E., Valverde, R. A., Kruse, L., Sabater, J. B., & Rodríguez-Molinero, A. (2018). Normal limits of home measured spatial gait parameters of the elderly population and their association with health variables. *Scientific Reports*, *8*(1), 13193. https://doi.org/10.1038/s41598-018-31507-1
8. OrthoFixar. (2022, May 1). *Gait Cycle | OrthoFixar 2023*. Orthofixar.com. https://orthofixar.com/basic-science/gait-cycle/#:~:text=Each%20stride%20is%20composed%20of
9. Wellmon, R. (2007). *Stride Width - an overview | ScienceDirect Topics*. Www.sciencedirect.com. https://www.sciencedirect.com/topics/nursing-and-health-professions/stride-width#:~:text=Normal%20stride%20width%20for%20adults
10. *Back pain - Symptoms and causes*. (n.d.). Mayo Clinic. Retrieved July 3, 2024, from https://www.mayoclinic.org/diseases-conditions/back-pain/symptoms-causes/syc-20369906?s=3#:~:text=Keep%20your%20feet%20shoulder%2Dwidth
11. *Average Shoulder Width and How to Measure Yours*. (n.d.). Healthline. https://www.healthline.com/health/average-shoulder-width#why-we-measure