TECHNICAL VALIDATION OF A BODY-WEIGHT CONTROLLED CLUTCH FOR ANKLE-FOOT ORTHOSES OF CHILDREN WITH CEREBRAL PALSY

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Abstract

Ankle-foot orthoses (AFOs) greatly improve gait in patients with Cerebral Palsy (CP). Some AFO designs allow for passive push-off support, however, these often limit the ankle's ROM during the swing phase of gait. This contribution presents the technical validation of a body-weight controlled clutch (BWC) designed for children with CP, to passively engage and disengage the push-off support without restricting ankle kinematics. We determined the friction coefficient (μ) of different BWC prototypes, and used it as an indicator for the amount off force that can be exerted on the mechanism before slippage occurs. Four clutch configurations were tested, containing a rigid or flexible spacer and a nylon strapping webbing or neoprene rubber slider. The best tested configuration was the one composed by the rigid spacer–nylon slider combination, which yielded a μ as high as 0.98. We envision that a lightweight solution like the BWC presented here can benefit new AFO designs to support push-off on children with gait deficiencies.

Keywords: Body-weight controlled clutch, Push-off, Gait, Cerebral Palsy

1 INTRODUCTION

Cerebral Palsy (CP) is a group of neurological disorders caused by permanent damage to the brain that happens before, during or shortly after childbirth [16, 19]. Many children with CP have problems with movement and coordination, and suffer from pathological gait patterns that progress over-time as the child grows [9, 2]. This causes increased energy cost of walking, pain, and joint degeneration [12, 17]. Current treatments to improve the walking ability of these patients include physical therapy, assistive devices, medication and surgery [11, 5].

Ankle-foot orthoses (AFOs) are the foremost used type of assistive devices for CP treatment due to the role of the ankle joint in gait [21, 10, 4]. Traditional AFO designs provide patients with CP with the required stability to improve their gait pattern. However, they present limitations in assisting gait, especially with regard to push-off power [13], which is normally associated with an increased energy cost of walking. In recent years, new assistive non-actuated AFO designs have been introduced, but they are mostly focused on assisting dorsiflexor muscles not plantarflexors to avoid drop-foot during the swing phase [1, 18]. There are only few advances for non-actuated AFO designs that support push-off on patients with motor disabilities, e.g. posterior leaf springs and adjustable dynamic response AFOs. These solutions implement passive springs that store energy during the stance phase of gait and subsequently release it, assisting push-off, similar to the functioning of the Achilles tendon.

Unfortunately, the spring modules of current non-actuated push-off assistive solutions do not act during the stance phase alone, but also present dorsiflexion stiffness, thereby potentially opposing toe clearance during swing [1]. Some approaches that attempt to solve this problem use clutches to activate their spring modules only during the stance phase, disconnecting them during swing [20, 22, 7, 15, 8, 14]. In this way, interference with swing phase kinematics is prevented, and the full desired range of motion (RoM) of the ankle is preserved.

One of these clutch-based approaches is the body-weight controlled clutch (BWC) [20, 22, 7, 15], a promising lightweight solution that allows clutching/unclutching based on the user's body-weight. However, BWCs are not yet commercially available, and most importantly, they have only been tested on healthy adults with the aim of enhancing human walking efficiency [20, 22, 7, 15].

The goal of this contribution is to design a BWC specifically for assisting push-off in children with CP. Various design configurations of this clutch are technically validated on a test bench in order to assess their performance and friction coefficients.

2 MATERIALS & METHODS

2.1 Design requirements

The design of the BWC is part of the research of inGAIT [3], an on going project that aims to enhance gait in children with CP by exploring possible improvements to current AFO technologies. Relevant requirements for the design of the BWC (Table 1) have been established taking into account those collected for the inGAIT.

2.2 Body-weight controlled clutch

We designed our BWC based on the concept presented by Yandell et al. [22], and adapted it to fit the requirements of our target population (i.e. children with CP, see section 2.1).

The BWC is placed underneath the user's foot and consists of a slider that can move in between two foot-sole shaped grippers (Figure 1). During the stance phase of gait, when the user bears weight on the sole, the normal force clutches the slider in between the two grippers. This blocks the attachment string, which will eventually be connected to an assistance spring at the back of the shank. Thus, energy will be stored within the assistance spring with the forward progression of the tibia during the stance phase. At the end of the stance phase, during pre-swing, the assistance spring pulls the ankle into plantarflexion, thereby assisting push-off. At toe off, the BWC is unloaded, releasing the slider and allowing the foot to move freely without feeling any resistance from the assistance spring. Finally, to ensure that the attachment string and assistance spring remain under tension, a reset spring is included at the front of the slider.

Two spacer prototypes of the BWC were created based on the shape of a child's sports shoe (EU-34 size): (1) a completely rigid spacer (3D printed PLA, 5 mm high); and (2) a flexible spacer (fast resetting foam, 9.5 mm high), see Figure 2. Both

Table 1: Design requirements considered within the inGAIT project. The RoM that the device allows at the ankle is considered to be negative towards dorsiflexion.

Requirement	Value
User's age	4 – 16 years
User's body-weight	15 – 60 kg
Length of the foot	15.8 – 25.3 cm
Allowable passive RoM	-15 – 20 degrees
Desired push-off assistance	0.3 Nm/kg
Mass of the mechanism	≤ 0.3 kg

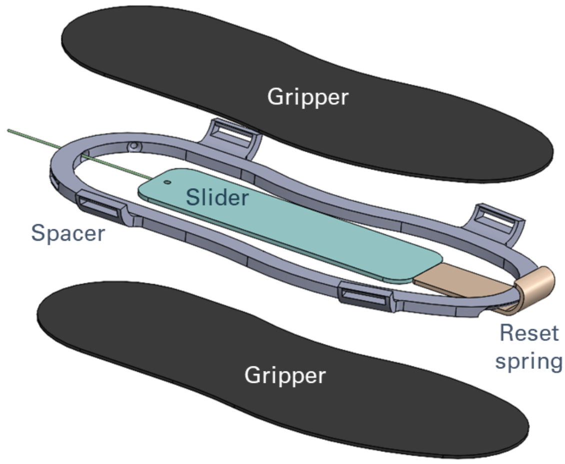

Figure 1: Clutch CAD design. It consists of a spacer that separates the top and bottom grippers, allowing the slider to move freely when no load is applied to the clutch. A reset spring keeps the attachment string connected to the slider under tension.

spacers had a ring width of 10 mm, with a 40 mm wide notch at the toes to pass the reset spring through, and a hole at the back of the heel for guiding the attachment string (Figures 1 and 2). To allow toe roll-off during walking, the rigid spacer was made with a 20 mm wide gap at the metatarsophalangeal joint.

Each spacer prototype could be fitted with one of two different sliders (50x130 mm), Figure 2: (1) a slider made of nylon strapping webbing (1.5 mm thick); and; (2) a slider made of neoprene rubber (2 mm thick).

2.3 Friction coefficient

To technically validate the different configurations of the BWC, it is important to assess the clutching properties of the two prototype sliders in combination with the two spacer prototypes.

Clutching efficacy depends on the friction coefficient, μ, between the slider and the grippers during clutch loading, which can be calculated with Equation 1:

$$\mu = \frac{F_{fric}}{F_N} \tag{1}$$

where, F_{fric} is the friction force, the maximum force that can be applied on the slider before it starts slipping, and F_N is the normal force acting on the clutch due to the user's body-weight.

The friction coefficient μ generally ranges from 0 to 1, where lower values indicate low clutching efficacy [6, 22]. In our prototype, we can

https://doi.org/10.17979/spudc.9788497498418.0165

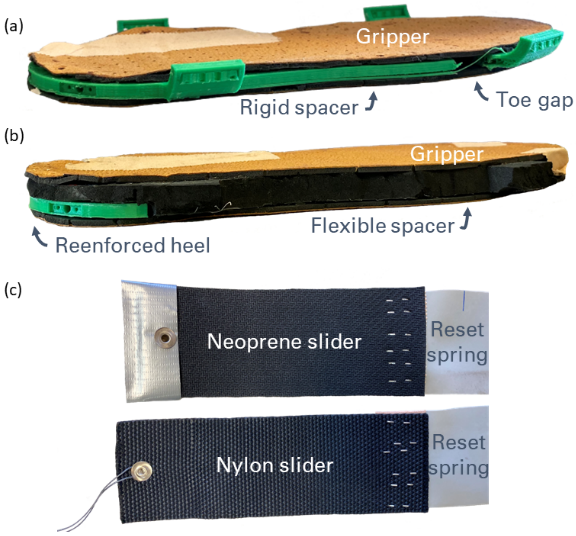

Figure 2: BWC prototypes: (a) rigid spacer with toe gap, (b) flexible spacer with rigid heel reenforcement, and (c) two different slider designs, (top) neoprene rubber and (bottom) nylon strapping webbing .

assess an effective friction coefficient which value depends not only on the slider and gripper materials, but also on the spacerÂ´s height relative to slider thickness. For example, a thinner spacer combined with a thicker slider will make it easier for the slider and gripper to form a firm connection, resulting in a higher effective μ. However, for proper unclutching during the swing phase, when the clutch is not loaded, it is also important that the space between the grippers is sufficiently large to allow free movement of the slider. For simplification, we will refer to this effective friction coefficient just as friction coefficient, μ.

Finally, it is worth noting that the friction coefficient does not depend on surface area. We assume Coulomb friction according to Equation 1 which does not contain the area [6]. However, a certain minimum area is required for the (maximum) friction coefficient of a system to be obtained.

2.4 Experimental setup and technical validation

A test bench was created for assessing the prototypes' μ coefficient for different normal force values (Figure 3). For each spacer–slider combination, the clutch was loaded with weights ranging from 15 to 45 kg in steps of 5 kg (Table 2). These weights were used to simulate the normal forces that the target users can exert on the clutch. Due to the diverse characteristics of patients with CP while walking, it is important to assess the clutching during different key events of the stance phase.

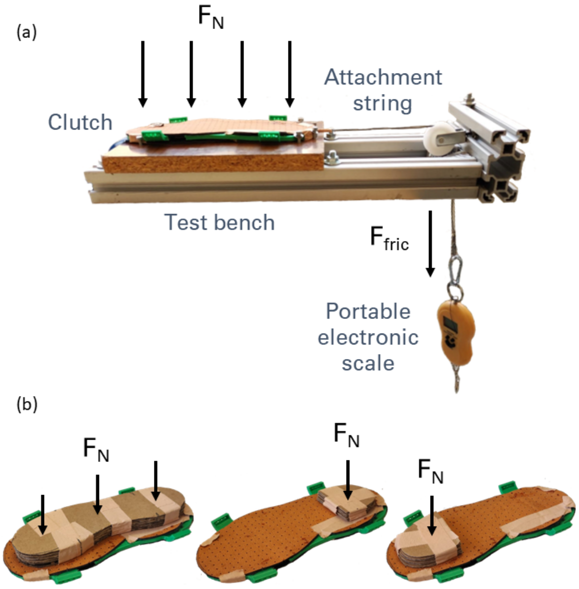

Figure 3: Experimental setup: (a) The clutch was fixed to the test bench and loaded with weights, simulating the F_N applied by the final users. The attachment string of the slider was guided through a pulley and connected to a portable electronic scale, which was used to measure the maximum F_{fric} until the slider slipped. (b) Different plateaus were placed on the clutch to ensure that the loading weight acted on the mid-foot, heel or toes.

Thereby, the different loading weights were tested at three locations (Table 2 and Figure 3): midfoot (to simulate mid-stance), heel (to simulate heel strike) and toes (to simulate push-off). Unlike other contributions [22], we also included the heel loading test as it would allow us to evaluate the potential of our BWC to start clutching immediately after heel strike, not delaying it to foot-flat.

To estimate μ for all parameter combinations in Table 2, we determined the maximum holding force (F_{fric}) in each case. The F_{fric} was obtained by applying pulling forces to the attachment string connected to the slider (Figure 3). This allowed us to simulate the force exerted on the slider by the assistance spring during the tibial progression of the stance phase. The pulling forces were manually exerted and measured via a portable electronic scale. They were progressively increased until the slider slipped (visually observed). The display of the portable electronic scale was filmed to determine the maximal manually applied holding force. Each measurement was executed three times. The F_{fric} was estimated by averaging the three registered values.

For each clutch configuration F_{fric} was plotted as a function of F_N, and a linear regression fit was performed. The resulting slopes were used to estimate the averaged μ for mid-foot, heel and toes loading. For the linear regression fits, as we did not have data points below 15 kg of normal weight, we assessed F_{fric} when no load (0 kg) was applied on the clutch, i.e. $F_N = 0\ N$, and used these values as intercept points. This allowed us to satisfy the physical constraint that F_{fric} cannot be negative when $F_N = 0\ N$.

Table 2: Overview of interchangeable experimental parameters

Parameters	Values
Spacer	Rigid, flexible
Slider	Nylon, neoprene
Normal weight (kg)	0, 15, 20, 25, 30, 35, 40, 45
Weight location	Mid-foot, heel, toes

3 RESULTS

The obtained values of μ for each clutch configuration are presented in Table 3. These correspond to the slopes of the linear regression fits for F_{fric} versus F_N datapoints (Figure 4).

The μ for mid-foot and heel loading are quite similar for all spacer-slider combinations (Table 3 and Figure 4), although the clutching surface for heel loading was much smaller. The slider was not long enough to properly reach the toe area and thus the μ was much lower for these cases.

For the mouse pad slider the μ at the toes was so low (0.01 and 0.00, for rigid and flexible spacer respectively) that it can be said that no clutching took place.

4 DISCUSSION

The main objective of this paper was the technical validation of a body-weight controlled clutch as a new feature for improving AFOs for children with CP. This was done by estimating the friction coefficient μ of the system for four clutch configurations, i.e. combinations of two spacers and two sliders, and three locations of F_N on the clutch: mid-foot, heel and toes.

Table 3: Obtained friction coefficients with RMS fit

Spacer	Slider	μ		
		mid	***heel***	***toe***
Rigid	Nylon	0.98	0.88	0.33
	Neoprene	0.48	0.42	0.01
Flexible	Nylon	0.54	0.54	0.25
	Neoprene	0.30	0.33	0.01

From all tested spacer–slider combinations, the highest clutching efficacy was obtained for the 5 mm rigid spacer combined with a nylon slider. Even though the flexible spacer was made from flexible fast resetting foam, it performed worse than the rigid spacer, as indicated by the lower μ. This can be explained by the fact that the flexible spacer was twice as thick as the rigid spacer, and thus made clutching more difficult.

The nylon slider performed better than the neoprene slider in combination with both the rigid and flexible spacer, as shown by the higher μ coefficient between gripper and nylon slider material than between gripper and neoprene slider material. Moreover, the nylon slider was able to withstand the forces that it was subjected to during the experiment, while the neoprene slider failed when loaded with 15 kg (e.g. rigid spacer with neoprene slider, Figure 4).

The slider of the final design should be able to carry the maximum user's weight of 60 kg and withstand the pulling forces applied by them. In that sense, the nylon slider seems to be strong enough to withstand this load, although we did not test it in this experiment, as we only reached up to 45 kg.

5 CONCLUSION

In this contribution we tested a BWC designed to passively engage and disengage the assistive push-off support of AFOs for children with CP. We envision that a lightweight solution like the one presented here can greatly benefit the current AFO designs for this population.

With our technical validation we showed that a sufficient friction coefficient can be reached with relatively cheap, lightweight and easy to find materials. However, the clutch validation is not complete yet. The next step should be to test the BWC with both, healthy users and patients with CP. That would provide some answers to the question if the mechanism does reduce energy cost of walking, while verifying that the clutching/unclutching is performed correctly and users do not feel any dorsiflexion stiffness during swing.

Acknowledgements

This work has been carried out with the financial support from the Dutch Research Council

https://doi.org/10.17979/spudc.9788497498418.0165

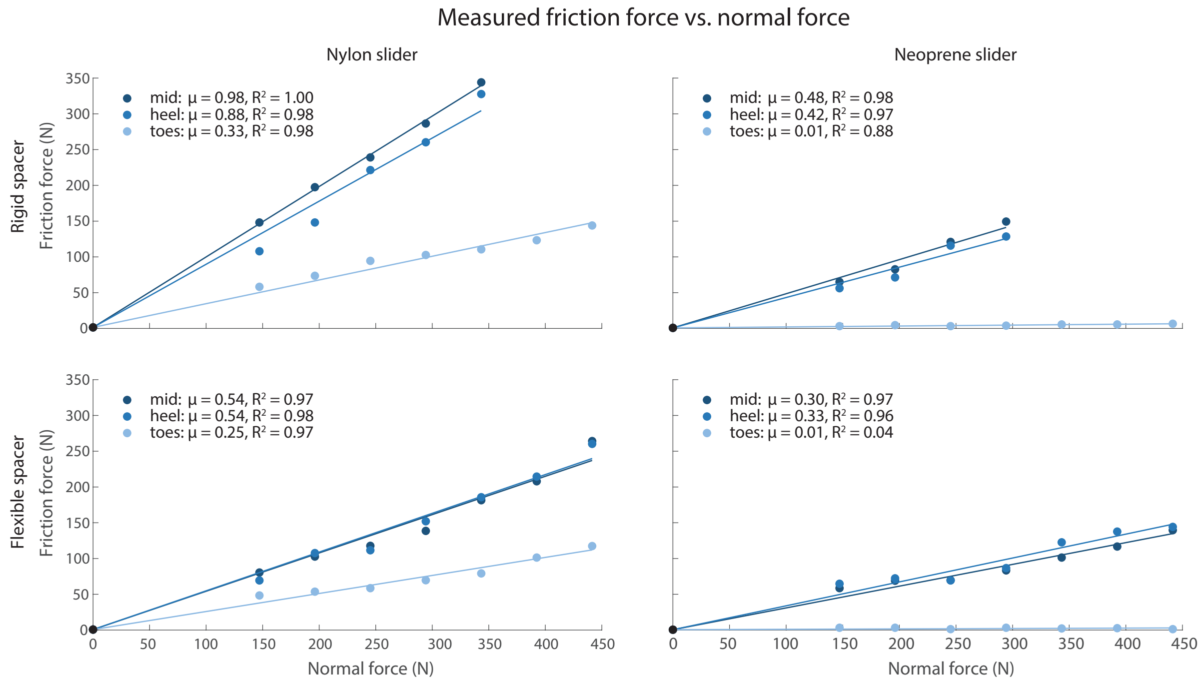

Figure 4: Obtained values when assessing the maximum F_{fric} pulling from the slider for a fixed F_N applied on the clutch. The position of F_N was tested at three locations (mid-foot, heel and toes). A linear fit was applied to the data points to find the corresponding μ and R^2 coefficients.

NWO, under the grant Veni-TTW-2020 with ref. number 18079, and partly with the financial support from the KNAW Ter Meulen grant of the Academy Medical Sciences Fund, Royal Netherlands Academy of Arts & Sciences, ref. number KNAWWF/1327/TMB202101.

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