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1 **Reliability outcomes and inter-limb differences in ankle joint stiffness in children with**
2 **unilateral cerebral palsy depend on the method of analysis**

3

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19

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21

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23

24

25

26

27 **1 Introduction**

28 Cerebral Palsy (CP) refers to motor control alterations after brain damage occurring
29 perinatally (Rosenbaum et al., 2007), and has a prevalence of 1.7 per 1000 live births in
30 Europe (Sellier et al., 2016). The most common form of CP is spastic CP resulting from
31 alterations to the pyramidal system that leads to a velocity-dependent increase in tonic stretch
32 reflexes with exaggerated tendon jerks (Lance 1980). Spastic CP is well known to induce
33 several functional motor impairments, especially in lower limb muscles, that may appear
34 during growth and alter quality of life. For instance, muscles contractures, defined a decrease
35 in muscle belly length (Lieber and Fridén, 2018) associated with an increase in passive
36 muscle stiffness (Farmer and James 2001), have been frequently reported for plantarflexor
37 (PF) muscles in spastic CP (Mathewson and Lieber, 2015, Švehlík et al., 2013). As a
38 consequence, ankle joint range of motion has been reported to decrease dramatically during
39 growth in children with CP (Hägglund and Wagner, 2011, Švehlík et al., 2013). Moreover,
40 due to the important functional role of the PF muscles in vertical support and forward
41 propulsion during walking (Perry et al., 1974), their alteration may induce gait impairments
42 such as equinus gait (Wren et al., 2010, Švehlík et al., 2010). Equinus foot deformity is one of
43 the most common problems leading to gait and functional limitations in children with CP
44 (Goldstein and Harper 2001, Švehlík et al., 2010, Wren et al., 2010).

45 Passive ankle joint stiffness (PAJS) has been classically assessed by establishing the torque-
46 angle relationship during passive dorsiflexion (DF) which may be considered as the ‘gold
47 standard’ for mechanical assessment of muscle-tendon properties (Moseley et al., 2001,
48 Harlaar et al., 2000, Tardieu et al., 1982). PAJS is then calculated as the slope of the torque-
49 angle relationship using a linear fit (Alhusaini et al., 2010, Barber et al., 2011, Kruse et al.,
50 2017, Ross et al., 2011, Theis et al., 2016). This has led to reports of increased PAJS in
51 children with CP when compared to typically developing (TD) peers (Alhusaini et al., 2010,

52 Barber et al., 2011, de Gooijer-van de Groep et al., 2013, Ross et al., 2011), although others
53 did not evidence differences between groups (Kruse et al., 2017, Ross et al., 2011). Moreover,
54 large discrepancies have been reported for the magnitude of increased stiffness in children
55 with CP among studies, ranging from 35% (Ross et al., 2011) to 225% (Alhusaini et al.,
56 2010). Such discrepancies could be due to differences in methodology and/or heterogeneity of
57 the sample of tested children.

58 First, the choice of the angular sector for slope calculation has been reported to largely
59 influence the results and their interpretation (Ross et al., 2011). Since all the aforementioned
60 studies used different angular [i.e. 0 to 5° for Alhusaini et al., (2010), -30 to -10° and last 20°
61 of DF for Ross et al., (2011)] or torque [30 to 100% for Barber et al., (2011), 20 to 80% for
62 Theis et al., (2015, 2016) and 50 to 90% of maximal passive torque for Kruse et al., (2017)]
63 sectors for slope calculation, comparison between studies may be hazardous and clear
64 recommendations should be provided to ensure accuracy of future studies. Here, we propose
65 to investigate the influence of PAJS calculation method on reliability outcomes and
66 interpretation that can be made from the results. We believe that this may help clinical
67 decision making such as to perform either conservative (e.g. casting, splinting, stretching) or
68 surgical treatment.

69 Secondly, most studies included children with both bilateral and unilateral CP (UCP)
70 notwithstanding that the rate of PF growth and strength differs among them (Barber et al.,
71 2016, Wiley and Damiano, 1998). Moreover, children with UCP are more prone to
72 developing fixed equinus deformity during growth because of a frequent combination of a
73 true equinus pattern (Goldstein et al., 2001, Winters et al., 1987) and dorsiflexors muscle
74 weakness (Cobeljic et al., 2009). Altogether, this may potentially influence passive stiffness.
75 Accordingly, it would be of interest to specifically investigate PAJS in children with UCP in a
76 clinical routine. Comparing the paretic and non-paretic legs could be of high interest for the

77 clinician in a longitudinal follow-up during growth and to evaluate the impact of therapeutic
78 strategies such as stretching program (Theis et al., 2015), home based exercises program
79 (Kruse et al., 2018) or gait training (Hösl et al., 2018; Willerslev-Olsen et al., 2015). Before
80 that, it is crucial to characterize the reliability of the measurement.

81 Therefore, the purpose of this study was to compare in children with UCP paretic and non-
82 paretic legs PAJS and to further determine the influence of different calculation methods on
83 the outcomes and their inter-session reliability. Because of differences in ankle range of
84 motion during passive stretch between both paretic and non-paretic legs, increased stiffness
85 may occur earlier in the range of absolute stretch for the spastic than non-spastic leg (Ross et
86 al., 2011). An additional aim of this study was to further investigate the inter-limb difference
87 when stiffness is calculated on the same absolute or relative angular sector.

88 **2 Method**

89
90 Thirteen children (7 females, 6 males) with spastic UCP (Table 1) participated in this study,
91 which was approved by the local ethics committee (CPP Sud-Est, number 2016-A01558-43)
92 and conducted according to the declaration of Helsinki. Parents and children gave their
93 written informed consent prior to inclusion in the study. Children (aged 5-12 years) with
94 spastic UCP (i.e. Tardieu scale $\geq X2$, V2) were recruited provided they were able to
95 understand and follow simple verbal instructions, had no botulinum toxin injections or cast
96 applications in the last three months, and had less than 10° of knee flexion contracture.
97 Children were excluded if they had any surgical procedures performed on the lower limbs.
98 Each child was tested twice at a 1-week interval and at the same time of the day by the same
99 experimental investigator. Paretic and non-paretic limbs were assessed in a random order in
100 the first session and the same order was kept for the second session.

101 After determining their Gross Motor Function Classification System (GMFCS) level,

102 spasticity level on soleus and gastrocnemius muscles on the paretic limb was assessed with
103 the Modified Ashworth and Tardieu Scales, and selective motor control was assessed with the
104 Boyd scale. **Passive** maximal DF angle was then measured manually for both limbs with the
105 knee extended and in supine position using a manual goniometer. After that, children were
106 installed in the prone position (with hip and knee at 0° extension (i.e. fully extended) and with
107 the ankle joint positioned **at 25° of ankle PF**) on an isokinetic dynamometer (Con-Trex®,
108 Medimex, France). The axis of the dynamometer was aligned with the presumed axis of
109 rotation of the ankle. The foot was firmly fixed to the dynamometer and the thigh was fixed to
110 the dynamometer with a Velcro™ attachment. **Experimental condition of ankle joint**
111 **positioning was presented Figure 1.** Torque-angle relationships were then recorded during
112 passive DF performed at 2°/s between **25°** of ankle PF and the pre-determined maximal DF
113 angle of the ankle. To ensure that muscles were totally relaxed during passive movements,
114 EMG activity was recorded using dry-surface electrodes (Delsys DE 2.1, Delsys Inc, Boston,
115 MA, USA) placed on the gastrocnemius lateralis and tibialis anterior muscles. Sampling
116 frequency for angle, torque and EMG signals was 2000 Hz. A total of 5 passive
117 loading/unloading cycles was performed to account for a possible conditioning effect
118 (Maganaris, 2003) and the last loading cycle (i.e. DF phase) was retained for joint stiffness
119 analysis since no significant changes have been reported to occur between the fourth and the
120 fifth cycles (Maisetti et al., 2012).

121 Considering the variety of methodological approach used in the literature to calculate PAJS,
122 stiffness was considered in the present study as the slope of the linear regression between
123 passive torque and ankle angle in the range between (1) 30 to 100% (method 1, Barber et al.,
124 2011), (2) 20 to 80% (method 2, Theis et al., 2015 and 2016) and (3) 50 to 90% of maximal
125 passive torque (method 3, Kruse et al., 2017) as previously performed by others. To account
126 for differences in range of motion between paretic and non-paretic legs, PAJS was calculated

127 for the non-paretic leg not only on the relative range defined above (PAJS_{rel}) but also on the
128 same absolute angular sector as for the paretic one (PAJS_{abs}).

129 **Statistical analysis**

130

131 Coefficients of determination (R^2) of the torque-angle relationships were calculated for linear
132 fitting. As recommended by Hopkins (Hopkins, 2000), inter session reliability of the PAJS
133 were assessed for each method (i.e. 30-100%, 20-80% and 50-90%) and both legs (i.e. non-
134 paretic leg and paretic leg on both the same absolute and relative angular sector as for the
135 paretic one) using the intra-class correlation coefficient (ICC), standard error of the mean
136 (SEM) and coefficient of variation (CV). The ICC provides a measure of relative reliability
137 showing the degree to which individuals maintain their position in a sample with repeated
138 measurements (Hopkins, 2000). According to Fleiss (Fleiss, 1981), $ICC > 0.75$ represents
139 excellent reliability; an ICC between 0.74 and 0.40 represents moderate to good reliability
140 and an $ICC < 0.40$ represents poor reliability. The SEM scores were also calculated to
141 quantify absolute reliability. Thus, a low SEM value means strong reliability and vice-versa
142 (Hopkins, 2000). The CV (standard deviation (SD) / mean) was calculated to represent the
143 degree to which repeated measurements vary for individuals (Hopkins, 2000). Reliability was
144 considered as good for $CV < 10\%$ and moderate for $CV 10-20\%$. All variables passed the
145 normality test (Kolmogorov–Smirnov normality test) and homogeneity of variance
146 verification (Levene’s test). A paired t-test was used to compare maximal DF angle between
147 paretic and non-paretic legs. Three-way ANOVAs were performed to determine the effect of
148 slope calculation method (i.e. method 1, 2 and 3), sessions (i.e. 1 and 2) and tested limb (i.e.
149 paretic leg, non-paretic leg, and non-paretic leg within the same absolute angular sector as for
150 the paretic leg) for PAJS as well as coefficients of determination (R^2) of the torque-angle
151 relationships. Two-ways ANOVAs were performed to determine the effect of slope

152 calculation method and tested limb for reliability outcomes (CV). When the ANOVA
153 identified significant effects, post-hoc Fisher testing was performed. Data are presented as
154 mean \pm SD. Statistical significance was set at $p < 0.05$.

155

156 **3 Results**

157 The characteristics of the population are shown in Table 1. Children with UCP presented
158 significantly decreased maximal ankle DF angle ($p < 0.0001$) (Table 1).

159 **Averaged torque-angle relationships obtained for the whole group during both first and**
160 **second testing sessions are presented for the paretic legs in Figure 2.**

161 Relationships were well fitted by the linear regression. Mean R^2 ranged between 0.94 (i.e.
162 paretic limb, session 1, 50-90%) and 0.97 (i.e. non-paretic leg within the same absolute
163 angular sector as for the paretic leg, session 2, 20-80%). There were no main effect of session
164 ($p = 0.3$) or tested limb ($p = 0.15$). However, there was a main effect of the calculation method
165 ($p < 0.001$) with post-hoc testing only revealing a trend for greater R^2 for method 2 (mean R^2 of
166 0.97) than method 3 (mean R^2 of 0.95) ($p = 0.077$). Moreover, there was a method \times tested
167 limb interaction effect ($p < 0.0001$). Post-hoc testing demonstrated higher R^2 value for method
168 2 (mean R^2 of 0.96) when compared to method 3 (mean R^2 of 0.94) on the paretic leg
169 ($p < 0.001$). Similarly, higher values were reported for method 1 and 2 (mean R^2 of 0.97 for
170 both methods) when compared to method 3 (mean R^2 of 0.94) on the non-paretic leg within
171 the same absolute angular sector as for the paretic leg ($p < 0.001$ and 0.0001, respectively).

172 **Using method 2, Figure 3 illustrates for the whole group the comparison of PAJS_{rel} and**
173 **PAJS_{abs} measurements between paretic and non-paretic legs.** Table 2 presents stiffness
174 measurement values as well as their inter-session reliability outcomes.

175 **When considering PAJS values, the three-ways ANOVA (limbs \times sessions \times methods)**
176 **revealed:**

177 - a significant main effect of method ($p < 0.0001$): post-hoc testing revealed significantly lower
178 PAJS values using method 2 (mean stiffness of 0.19 Nm°) when compared to method 1
179 (mean stiffness of 0.23 Nm°) ($p = 0.009$) and to method 3 (mean stiffness of 0.26 Nm°)
180 ($p < 0.0001$) while the difference was almost significant between the latter two ($p = 0.06$).

181 - a significant method \times tested limb interaction effect ($p = 0.009$): post-hoc testing revealed that
182 stiffness values were significantly higher for the paretic leg when compared to non-paretic
183 PAJSabs whatever the method used for calculation ($p < 0.001$ with method 1, $p = 0.007$ with
184 method 2, $p < 0.0001$ with method 3). No significant difference was found between paretic leg
185 and non-paretic leg PAJSrel values whatever the method used for calculation ($p = 0.69$, 0.86
186 and 0.40 with method 1, 2 and 3, respectively).

187 - no interaction effect involving the factor session (session \times method: $p = 0.56$; sessions \times limbs
188 \times methods: $p = 0.42$).

189 Regarding the inter-session CVs, two-ways ANOVA revealed significant main effect of
190 method ($p = 0.001$). Post-hoc testing demonstrated significantly higher CV values for method 3
191 (mean CV of 11.7%) when compared to method 2 (mean CV of 6.6%) ($p = 0.03$), and a non-
192 significant difference when compared to method 1 (mean CV of 7.5%) ($p = 0.07$). CVs were
193 not different between methods 1 and 2 ($p = 0.69$). Two-ways ANOVA also revealed a
194 significant main effect of the tested limb ($p < 0.001$). Higher CV values were reported for the
195 paretic limb (mean CV of 13.8%) when compared to the non-paretic one (mean CV of 7%)
196 ($p = 0.04$) and to the non-paretic leg within the same absolute angular sector as for the paretic
197 leg (mean CV of 5%) ($p = 0.01$). There was no difference between CV values of the latter two
198 ($p = 0.53$). Finally, there was no significant method \times tested limb interaction effect ($p = 0.08$).

199 **4 Discussion**

200
201 In the present study, we investigated in children with UCP inter-limb difference in PAJS
202 through torque-angle curve measurements. We further investigated how the methodological

203 approach (i.e. the selected angular sector for stiffness calculation) may influence PAJS values
204 of both limbs as well as their reliability outcomes. Our results suggest that the three tested
205 method for slope calculation (i.e. 30-100%, 20-80% and 50-90% of maximal passive torque
206 for methods 1, 2 and 3, respectively) led to similar results. For instance, there was no inter-
207 limb difference when considering $PAJS_{rel}$ while values were greater for the paretic limb when
208 compared to non-paretic $PAJS_{abs}$. Moreover, measurements were reported to be reliable,
209 although lower reliability was observed for the third method as well as for the paretic limb.

210 All the tested method for slope calculation were well fitted by a linear fit, whatever the
211 considered limb (R^2 ranged from 0.94 to 0.97). This confirms that linear regression is
212 appropriate to calculate PAJS as the slope of the torque-angle curve. Yet, method 3 was
213 significantly less fitted by the linear regression. Moreover, we report mean PAJS values in
214 agreement with previous reports in children with CP with similar characteristics than in the
215 present study (Ross et al., 2011).

216 Based on CV and ICC outcomes, PAJS reliability was considered as good for the non-paretic
217 leg whatever the method used for calculation (mean CV=7% and 5%, mean ICC=0.84 and
218 0.86 for $PAJS_{rel}$ and $PAJS_{abs}$, respectively) and moderate for the paretic leg (mean CV=13.8%
219 and mean ICC=0.62). For instance, we reported significant **higher** CV values for the paretic
220 leg, in agreement with findings on adults with hemiplegia (Harlaar et al., 2000). While this
221 may be the result of spasticity (Harlaar et al., 2000), we used a very low angular velocity
222 ($2^\circ/s$) to avoid reflex activation. While the reason for such difference in reliability between
223 limbs remains unknown, this may be the consequence of misalignment of the axes of the
224 ankle and dynamometer and/or motion of the knee during testing (Barber et al., 2011). These
225 methodological considerations may have predominantly affected measurements on the paretic
226 leg because of its likely smaller Achilles tendon moment arm (Kalkman et al., 2017) and foot
227 size. Consequently, correct ankle joint positioning and correct alignment of the dynamometer

228 were complicated on the paretic leg. Nonetheless, despite lower reliability was found on the
229 paretic leg, we reported CV values (12.4%, 9.7% and 19.2% for method 1, 2 and 3
230 respectively, table 1) largely lower than the decreased ankle joint stiffness reported after
231 interventions such as long-term stretching program (-32%, Theis et al., 2015), or gait training
232 (~40%, Willerslev-Olsen et al., 2015). This suggests that performing such measurements in
233 the clinical field may still make sense to evaluate the benefits of various treatment strategies.
234 Our results also demonstrated good inter-session reliability for methods 1 (mean value for
235 CV=7.5% and ICC= 0.78) and 2 (mean value for CV=6.6% and ICC= 0.79) for all the tested
236 limbs, as it was for method 3 on the non-paretic leg when considering both the absolute or
237 relative angular sector (mean value for CV=8% and ICC= 0.82). Conversely, lower and
238 moderate reliability was found when using method 3 on the paretic leg (CV=19.2% and
239 ICC=0.6).

240 Using a lower limit of 20% of maximal torque is thought to avoid the toe region at the lower
241 end of the stiffness curve (Abellaneda et al., 2009). As such, the three tested methods should
242 not have been influenced by this phenomenon. Moreover, an upper limit of 80% of maximal
243 torque is thought to minimize the contribution of passive elastic structures such as ligaments,
244 connectives tissues and skins at the upper end of the curve (Abelladena et al., 2009). EMG
245 responses of PF muscles may also tend to rise significantly from 80% of maximal torque
246 (McNair et al., 2001). Nonetheless, these influences may not explain the lower reliability
247 observed for method 3 (50-90% of maximal torque) because reliability was still good for
248 method 1 calculating slope of the torque-angle curve up to 100% of maximal torque.

249 Alternatively, we suggest that the lower reliability may be the result of the restricted range of
250 torque (i.e. 40% for method 3 vs 70% and 60% for methods 1 and 2, respectively) for slope
251 calculation that may be more sensitive to inter-session variability, especially for the paretic
252 leg. **Taking together, we suggest to choose the method 2 (i.e. 20-80% of maximal torque)**

253 given that our findings present better reliability outcomes for the paretic leg and avoid the
254 final portion of the torque-angle curve (i.e. last 20% of maximal torque).

255 To the best of our knowledge, this study is the first to compare PAJS between paretic and
256 non-paretic legs in children with UCP. Whatever the method used for slope calculation, we
257 found no difference in PAJS between paretic and non-paretic legs when considering the same
258 relative angular sector. While this is to our knowledge the first study to specifically compare
259 both legs in children with UCP, this finding is in agreement with a recent study of Kruse et
260 al., (2017) demonstrating using the third method we presented no difference in PAJS between
261 children (mean age: 11.9 years) with CP and TD peers. Conversely, Barber et al., (2011)
262 reported significant differences between CP young adults with CP (mean age: 18 years) and
263 TD peers using method 1. Discrepancies between studies may likely be the consequence of
264 differences in clinical status (e.g. CP severity) and age of participants. In the present study,
265 lack of inter-limb difference in PAJS despite reduced range of motion in the paretic side is in
266 agreement with the notion that non-paretic leg in children with UCP cannot be considered as
267 an “healthy” leg (Wiley et al., 2008). It has accordingly been reported in hemiparetic post-
268 stroke adults no inter-limb difference in PAJS while the paretic leg presented significantly
269 higher passive stiffness when compared to TD adults (Freire et al., 2017). Moreover, this is
270 consistent with findings of altered structural properties observed on both paretic and non-
271 paretic medial gastrocnemius muscles in children with UCP when compared to TD Peers
272 (Obst et al., 2017).

273 When taking into account inter-limb difference in maximal DF angle (i.e. by calculating slope
274 of the torque-angle curves on a same absolute angular sector for both legs), PAJS was
275 significantly increased in the paretic leg when compared to contralateral one whatever the
276 method used. Our results demonstrated 37%, 28% and 45% inter-limb differences for method
277 1, 2 and 3 respectively. We assume that comparing both legs on the same angular sector on

278 legs with significantly different maximal DF angle led to a comparison of different portions of
279 the torque-angle curve. Consequently, the slope of the curve should be systematically higher
280 on the limb with significantly lower maximal DF angle (Ross et al., 2011). Nonetheless, this
281 method allowed us to quantify increased resistance to stretch of the PF muscle-tendon unit of
282 the paretic leg when compared to the non-paretic one, just as the clinician would perceive it
283 while manually mobilizing the ankle joint of both legs through the same angular range (Ross
284 et al., 2011). This increased stretch resistance at a given ankle angle is further supported by
285 the higher gastrocnemius medialis stiffness reported on the paretic leg when compared to the
286 contralateral one at neutral angle (i.e. 0°) using Shear Wave Elastography in children with
287 UCP (Lee et al., 2016). Such an inter-limb difference in PAJS could be due to a mismatch
288 between bone and muscle length development in the paretic leg during growth (Pingel et al.,
289 2017), with decreased gastrocnemius muscle length (Malaiya et al., 2007). Although PAJS
290 does not necessarily reflect muscular (i.e. gastrocnemius and soleus) or tendinous (i.e.
291 Achilles Tendon) tissues behaviors during passive mobilization in children with CP (Bar-On
292 et al., 2018), musculo-tendinous unit, and especially muscle belly in children with CP, is
293 considered as the primary contributor to joint passive stiffness (Blazevich, 2018, Lieber et al.,
294 2017). Moreover, PAJS can be reliably measured using a simple dynamometer, as confirmed
295 by the present study, while measurements at muscle and/or tendon level would require a
296 complex experimentation design using ultrasonography in addition to dynamometry.
297 Accordingly, PAJS may be measured in a clinical routine as a valuable and convenient
298 parameter (1) to give useful objective parameters for clinical follow-up of children with CP
299 during growth, and (2) to evaluate the impact of therapeutic strategies. This may be especially
300 relevant for children with UCP who are more likely to develop ankle contracture and/or to be
301 less responsive to botulinum toxin injection than children with bilateral CP (Hastings-Ison et
302 al., 2016).

303 **5 Conclusions**

304 The present results suggest that the non-paretic leg in children with UCP may serve as a
305 “control” limb during the clinical follow-up, provided that the comparison with the paretic leg
306 is performed on the same absolute angular sector. We also suggest that PAJS should be
307 assessed using method 2, i.e. 20-80% of maximal passive torque.

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30° PF

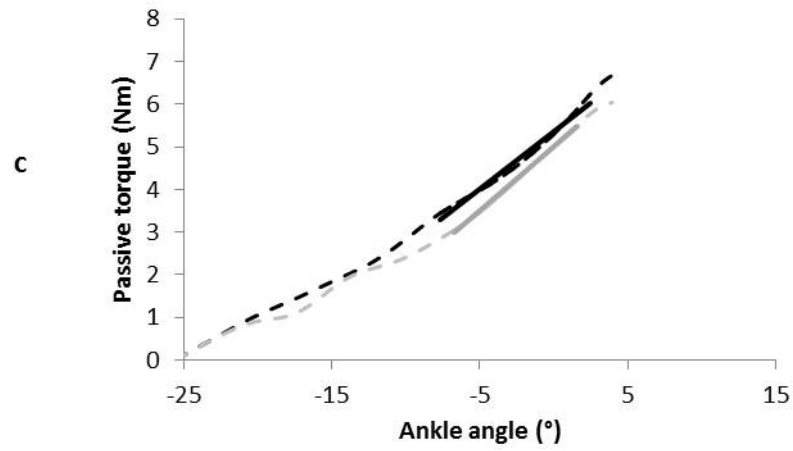
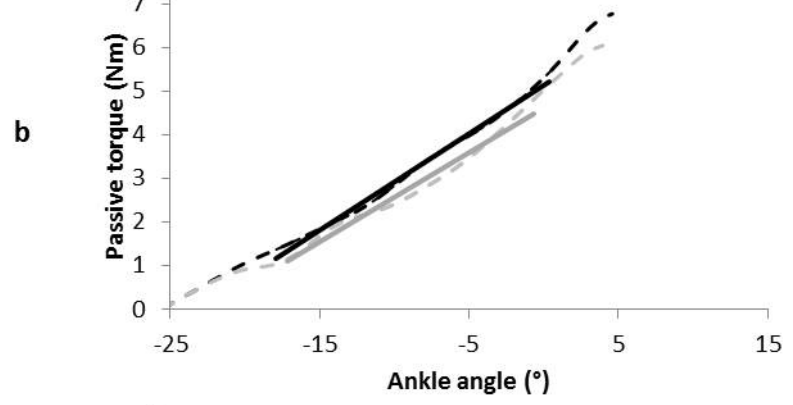
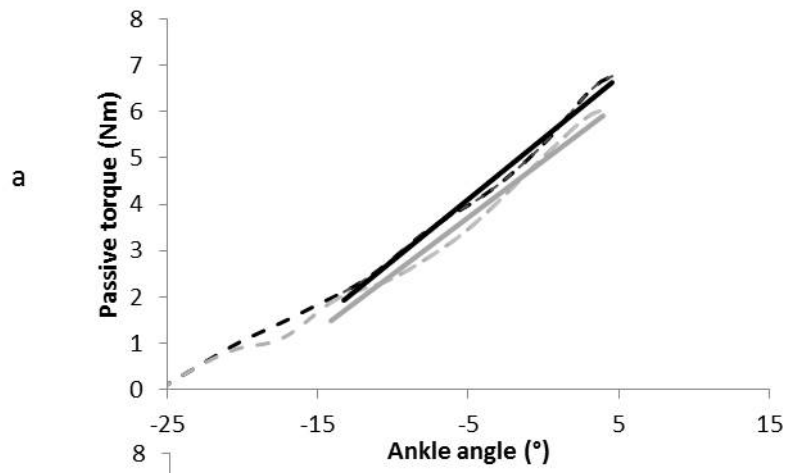
0°
(neutral angle)

15° DF

Plantarflexion



Dorsiflexion



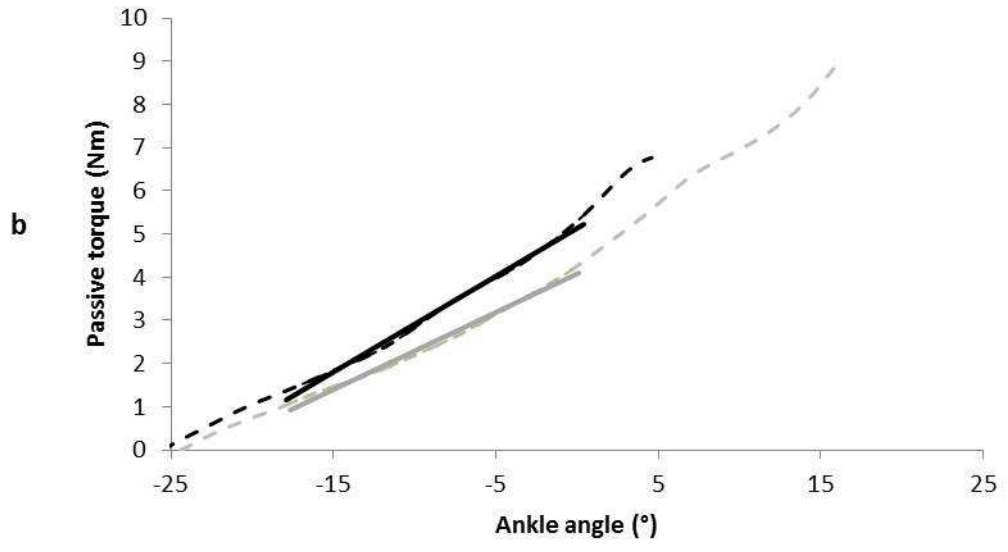
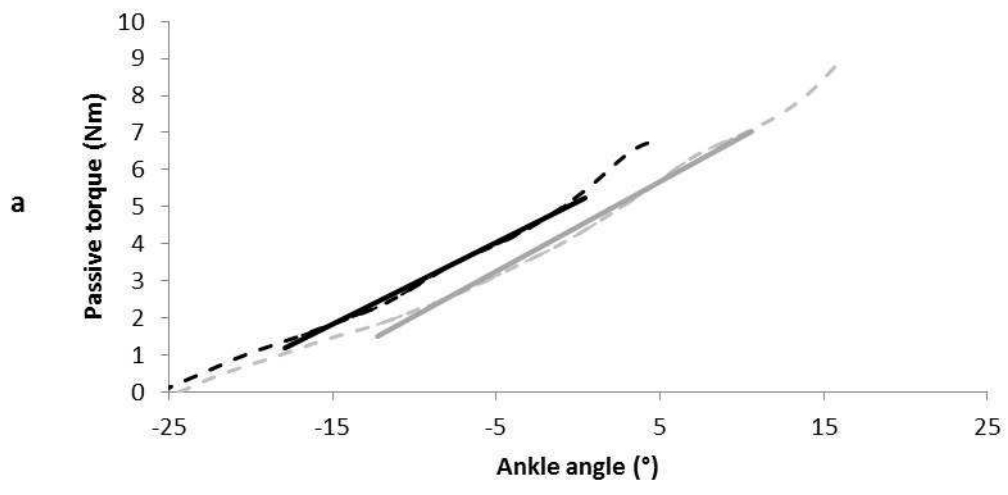


Table 1: General characteristics of the tested group; Data are presented as mean (SD) (*Nb*: number, *GMFCS*: Gross Motor Function Classification System, *DF*: dorsiflexion).

n	13
Sex (M/F)	6/7
Paretic side (right/left)	6/7
Age (Y:Mo)	9.8 (1.8)
Weight (kg)	32.6 (9.6)
Height (cm)	137.8 (13.5)
Tibia length on paretic leg (cm)	32.9 (4)
Tibia length on non-paretic leg (cm)	33.9 (4.1)
GMFCS Level, n	I; n=13
Modified Ashworth Scale	2.46 (0.77)
Tardieu Scale	1.2 (0.86)
Boyd scale	1.23 (0.44)
Maximum DF angle paretic side (°)	2.1 (6.5)
Maximum DF angle non-paretic side (°)	12.7 (3.8)
Nb Botulinum toxin injection	6.5 (2)
Nb weeks ankle casting	6.5 (5.6)
Nb session physiotherapy per week	1.5 (0.6)
Night splinting (Yes/No)	11/2
Gait orthoses (Yes/No)	10/3

Table 2: inter-session reliability outcomes for passive ankle joint stiffness (PAJS, in Nm/°) on paretic leg (PAJS_{rel}) and non-paretic leg on its relative (PAJS_{rel}) and absolute (PAJS_{abs}) angular sector calculated with three different methods. (*ICC: intra-class correlation coefficient, CV: coefficient of variation, SEM: standard error of measurement*).

Inter-session reliability		Mean (SD)				
Method	Side leg	Session 1	Session 2	ICC	CV (%)	SEM
Method 1	PAJS _{rel} paretic leg	0.24 (0.09) [°]	0.26 (0.06) [°]	0.57	12.41 ^ψ	0.05
	PAJS _{rel} non-paretic leg	0.25 (0.05)	0.25 (0.06)	0.84	6.5	0.02
	PAJS _{abs} non-paretic leg	0.18 (0.003)	0.18 (0.03)	0.93	3.48	0.01
Method 2	PAJS _{rel} paretic leg	0.20 (0.06) ^{α°}	0.21 (0.04) ^{α°}	0.69	9.72 ^ψ	0.03
	PAJS _{rel} non-paretic leg	0.20 (0.03) ^α	0.20 (0.04) ^α	0.86	5.36	0.01
	PAJS _{abs} non-paretic leg	0.15 (0.02) ^α	0.16 (0.02) ^α	0.83	4.59	0.01
Method 3	PAJS _{rel} paretic leg	0.26 (0.13) [°]	0.30 (0.11) [°]	0.6	19.21* ^ψ	0.07
	PAJS _{rel} non-paretic leg	0.29 (0.06)	0.30 (0.08)	0.82	9.15*	0.03
	PAJS _{abs} non-paretic leg	0.19 (0.06)	0.19 (0.04)	0.82	6.84*	0.02

* Significant higher CV compared to method 2

^ψ Significant higher CV compared to PAJS_{rel} non-paretic leg and PAJS_{abs} non-paretic leg

^α Significant lower PAJS values compared to method 1 and 3

[°] Significant higher PAJS_{rel} on the paretic leg compared to PAJS_{abs} on the on-paretic leg