Biomechanical measures of short-term maximal cycling on an ergometer: a test-retest study.

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Biomechanical measures of short-term maximal cycling on an ergometer: a test-retest study

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1 Biomechanical measures of short-term maximal cycling on an

2 ergometer: a test-retest study

3　　　　An understanding of test-retest reliability is important for biomechanists, such

4　　　　as when assessing the longitudinal effect of training or equipment interventions.

5　　　　Our aim was to quantify the test-retest reliability of biomechanical variables

6　　　　measured during short-term maximal cycling. Fourteen track sprint cyclists

7　　　　performed 3 x 4 s seated sprints at 135 rpm on an isokinetic ergometer,

8　　　　repeating the session 7.6 ± 2.5 days later. Joint moments were calculated via

9　　　　inverse dynamics, using pedal forces and limb kinematics. EMG activity was

10　　　measured for 9 lower limb muscles. Reliability was explored by quantifying

11　　　systematic and random differences within- and between-session. Within-session

12　　　reliability was better than between-sessions reliability. The test-retest reliability

13　　　level was typically moderate to excellent for the biomechanical variables that

14　　　describe maximal cycling. However, some variables, such as peak knee flexion

15　　　moment and maximum hip joint power, demonstrated lower reliability,

16　　　indicating that care needs to be taken when using these variables to evaluate

17　　　biomechanical changes. Although measurement error (instrumentation error,

18　　　anatomical marker misplacement, soft tissue artefacts) can explain some of our

19　　　reliability observations, we speculate that biological variability may also be a

20　　　contributor to the lower repeatability observed in several variables including

21　　　ineffective crank force, ankle kinematics and hamstring muscles' activation

22　　　patterns.

23　　　Keywords: sprint cycling, kinematics, kinetics, emg, maximal power.

Introduction

25 The reliability of a clinical or sports science test is defined as the consistency or

26 reproducibility of a performance when a test is performed repeatedly (Hopkins, Schabort, &

27 Hawley, 2001). This is an important consideration for researchers, clinicians and applied

28 sports scientists as the better the reliability of the measurement the easier it is to detect a real

29 change in outcome (Hopkins, 2000). If the reliability of a test is low, then the outcome of a

30 test may conceal the true effect of an intervention. Conversely, if the reliability of a test is not

31 known then small random deviations may be misinterpreted as a meaningful change in

32 performance (Yavuzer, Öken, Elhan, & Stam, 2008).

33 Applied biomechanics researchers are often interested in assessing the short- or long-term

34 effects of interventions that aim to improve clinical or sports performance outcomes. In

35 clinical gait analysis, for example, the results of biomechanical assessments are used to

36 inform clinical decision making, by evaluating the effectiveness of interventions such as

37 surgery, physical therapy, medication or orthotics on gait biomechanics (Kadaba et al., 1989;

38 McGinley, Baker, Wolfe, & Morris, 2009; Yavuzer et al., 2008). Test-retest reliability studies

39 of clinical gait have found that the sagittal plane kinematics and kinetics have high values of

40 reliability in comparison to the data collected in the transverse and coronal planes (McGinley

41 et al., 2009). Furthermore, knee abduction/adduction and hip, knee and foot rotation joint

42 angles demonstrate the lowest reliability (McGinley et al., 2009), with the size of the

43 measurement error the same order of magnitude as the real joint motion in these planes. In

44 the context of clinical gait therefore, reliability studies have proved valuable by identifying

45 those variables that need to be interpreted with particular caution in order to effectively

46 inform clinical decision making (McGinley et al., 2009).

47 An understanding of test-retest reliability has similar relevance when assessing sporting

48 movements, as biomechanical measures are often used to evaluate the effectiveness of

49 longitudinal interventions such as changes to training programmes or equipment modification

50 (Costa, Bragada, Marinho, Silva, & Barbosa, 2012; Milner, Westlake, & Tate, 2011). Cycling

51 is a commonly used sporting movement for this purpose, as it is a relatively constrained

52   movement that can be accurately manipulated (Neptune, Kautz, & Hull, 1997; Neptune &

53   Kautz, 2001). Whilst the reliability of submaximal or "endurance" cycling is well reported

54   (Bini & Hume, 2013; Hopkins et al., 2001; Jobson, Hopker, Arkesteijn, & Passfield, 2013;

55   Laplaud, Hug, & Grélot, 2006), only a small amount by comparison is known about the

56   reliability of short-term maximal cycling. This comparative deficit exists despite maximal

57   cycling being an important paradigm for studying physiological capacity (Coso & Mora-

58   Rodríguez, 2006), muscle coordination and motor control strategies, as well as having direct

59   relevance to a range of competitive cycling performance environments (Martin, Davidson, &

60   Pardyjak, 2007). Therefore, quantifying test-retest reliability in maximal cycling

61   biomechanics is important. Test-retest reliability has been quantified for overall net crank

62   power output on an inertial load cycling ergometer within- and between-session (Coso &

63   Mora-Rodríguez, 2006; Hopkins et al., 2001; Mendez-Villanueva, Bishop, & Hamer, 2007),

64   with trained cyclists producing reliable power within the first testing session (Martin,

65   Diedrich, & Coyle, 2000). These studies demonstrated within-session reliability was better

66   than between-sessions reliability for overall net crank power output (Coso & Mora-

67   Rodríguez, 2006; Martin et al., 2000). There have been no studies quantifying the within- and

68   between-session reliability of biomechanical variables (crank power and forces, joint angles,

69   angular velocities, moments and powers and EMG activity) for short-term maximal cycling

70   despite these measures being important descriptors of the outcome, technique and

71   intermuscular coordination of a movement (Brochner Nielsen et al., 2018; Jacobs & van

72   Ingen Schenau, 1992; Wakeling, Blake, & Chan, 2010). EMG activity can be used to

73   determine muscle activation onset and offset times and level of activation (Dorel, Guilhem,

74   Couturier, & Hug, 2012; Hug & Dorel, 2009). This is important when investigating

75   intermuscular coordination in cycling as the timing and magnitude of muscle activation has to

76   be coordinated appropriately to allow an efficient energy transfer from the muscles though

77   the body segments to the pedal (Neptune & Kautz, 2001; Raasch, Zajac, Ma, & Levine,

78   1997). Joint kinetic measures (moments and powers) at the hip, knee and ankle throughout

79   the pedal revolution describe the action and contribution of the joints to pedal power and can

80   be used to identify different coordination strategies between cyclists (Elmer, Barratt, Korff, &

81   Martin, 2011; Martin & Brown, 2009; McDaniel, Behjani, Brown, & Martin, 2014).

82   Combining information on muscle activation from EMG and joint kinetics from inverse

83   dynamics analysis provides a deeper understanding of the joint and muscle actions that

84   produce the movement, and hence both are required to describe intermuscular coordination in

85   maximal cycling and were chosen for measurement and analysis during maximal cycling

86   (Brochner Nielsen et al., 2018; Dorel, 2018).

87   The aim of this study was to quantify the test-retest reliability of kinematic, kinetic, and

88   muscle activation variables during maximal sprint cycling. We hypothesise that within-

89   session reliability would be better than between-sessions reliability.

90   **Methods**

91   *Participants*

92   Fourteen track sprint cyclists participated in the study. Participants regularly competed at

93   track cycling competitions at either Master's international and national level (10), or Junior

94   national level (4). Although the participants were varied in their anthropometrics (7 males

95   and 7 females, age: 40.5 ± 17.7 yr, body mass: 72.5 ± 8.5 kg, height: 1.71 ± 0.06 m,), they

96   were similar with respect to cycling performance level (flying 200 m personal best: $11.98 \pm$

97   0.90 s). Participants were provided with study details and gave written informed consent. The

98   study was approved by the Sheffield Hallam University Faculty of Health and Wellbeing

99   Research Ethics Sub-Committee.

Experimental protocol

An isokinetic ergometer was set up to replicate each participant's track bicycle position. All participants' crank lengths were set to 165 mm, which was what they rode on their track bicycles. Riders undertook their typical warm-up on the ergometer at self-selected pedalling rate and resistance for at least 10 minutes, followed by one familiarisation sprint (4 s at 135 rpm). Martin and colleagues demonstrated that trained cyclists can produce valid and reliable results for maximal cycling power from the first testing session (Martin et al., 2000), therefore one familiarisation sprint was deemed appropriate. Riders then conducted 3 x 4 s seated sprints at a pedalling rate of 135 rpm on the isokinetic ergometer with 4 minutes recovery between efforts. Participants undertook an identical session 7.6 ± 2.5 days apart, at approximately the same time of day (0.11 ± 2.18 h). A pedalling rate of 135 rpm was chosen as this is a typical pedalling rate during the flying 200 m event in track cycling and within the optimal pedalling rate range for track sprint cyclists (Dorel et al., 2005). The competitive level and typical training volume of our participants meant that it was not feasible to ask them to stop exercising 24 hours prior to the testing sessions, so instead they were instructed to undertake the same training in the preceding 24 hours before both sessions.

Isokinetic ergometer

A SRM Ergometer (Julich, Germany) cycle ergometer frame and flywheel were used to construct an isokinetic ergometer. The modified ergometer flywheel was driven by a 2.2-kW AC induction motor (ABB Ltd, Warrington, UK). The motor was controlled by a frequency inverter equipped with a braking resistor (Model: Altivar ATV312 HU22, Schneider Electric Ltd, London, UK). This set-up enabled the participants to start their bouts at the target pedalling rate, rather than expending energy in accelerating the flywheel. The ergometer was fitted with Sensix force pedals (Model ICS4, Sensix, Poitiers, France) and a crank encoder

124   (Model LM13, RLS, Komenda, Slovenia), sampling data at 200 Hz. Normal and tangential

125   pedal forces were resolved using the crank and pedal angles into the effective (propulsive)

126   and ineffective (applied along the crank) crank forces (Figure 1).

127   *Kinematic and Kinetic Data Acquisition*

128   Two-dimensional kinematic data of the participants' left side were recorded at 100 Hz using

129   one high speed camera with infra-red ring lights (Model: UI-522xRE-M, IDS, Obersulm,

130   Germany). The camera was perpendicular to the participant, centred on the ergometer and set

131   about 3 m from the ergometer. The camera was in a very similar position for both sessions.

132   Reflective markers were placed on the pedal spindle, lateral malleolus, lateral femoral

133   condyle, greater trochanter and iliac crest. The same researcher attached the markers for all

134   sessions. Kinematics and kinetics on the ergometer were recorded by CrankCam software

135   (Centre for Sports Engineering Research, SHU, Sheffield, UK), which synchronised the

136   camera and pedal force data (down sampled to 100 Hz to match the camera data) and was

137   used for data processing, including auto-tracking of the marker positions.

138   *EMG Data Acquisition*

139   EMG signals were recorded continuously from nine muscles of the left leg: vastus lateralis

140   (VL), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), long head of biceps

141   femoris (BF), semitendinosus (ST), lateralis gastrocnemius (GL), soleus (SO), and gluteus

142   maximus (GMAX) with Delsys Trigno wireless surface EMG sensors (Delsys Inc, Boston,

143   MA). The skin at electrode placement sites was prepared by shaving the area then cleaning it

144   with an alcohol wipe. The EMG sensors were then placed in the centre of the muscle belly -

145   with the bar electrodes perpendicular to the muscle fibre orientation, using the guidelines in

146   (Konrad, 2005) and secured using wraps to reduce motion artefacts during pedalling. The

147   same researcher attached the EMG sensors for all sessions. A Delsys wireless sensor

148   containing an accelerometer (148 Hz sampling rate) was attached to the left crank arm to

149   obtain a measure of crank angle synchronised with the EMG signals. The EMG system was

150   operated and recorded in EMGworks Acquisition software (Delsys Inc, Boston, MA),

151   sampling data at 1926 Hz. The Delsys trigno EMG system automatically applied a bandwidth

152   filter of 20 ± 5 Hz to 450 ± 50 Hz (>80 dB/dec) to the raw signals.

153   *Data Processing*

154   All kinetic and kinematic data were filtered using a Butterworth fourth order (zero-lag) low

155   pass filter with a cut off frequency of 14 Hz selected using residual analysis (Winter, 2009).

156   The same cut off frequency was chosen for the kinematic and kinetic data as recommended

157   by Bezodis and colleagues to avoid data processing artefacts in the calculated joint moments

158   (Bezodis, Salo, & Trewartha, 2013). Instantaneous crank power was calculated from the

159   product of the left crank torque and the crank angular velocity. The average left side crank

160   power was calculated by averaging the instantaneous crank power over a complete pedal

161   revolution. Owing to a technical fault with the force measurement in the right pedal, it was

162   not possible to calculate total average crank power per revolution (sum of left and right crank

163   powers). Joint angles were calculated using the convention shown in Figure 1. Joint moments

164   were calculated via inverse dynamics (Elftman, 1939), using pedal forces, limb kinematics,

165   and body segment parameters (de Leva, 1996). Joint extension moments were defined as

166   positive and joint flexion moments as negative. The joint moments are presented from the

167   internal perspective (Derrick et al., 2020). Joint powers at the ankle, knee and hip were

168   determined by taking the product of the net joint moment and joint angular velocity.

169   Insert Figure 1

170   Data were analysed using a custom Matlab (R2017a, MathWorks, Cambridge, UK) script.

171   Each sprint lasted for 4 s providing six complete crank revolutions which were resampled to

172   100 data points around the crank cycle. Crank forces and powers, joint angles, angular

173   velocities, moments and powers were averaged over these revolutions to obtain a single

174   ensemble-averaged time series for each trial.

175   The accelerometer data for the crank arm was filtered using a Butterworth fourth order low

176   pass filter with a cut off frequency of 10 Hz. The minimum value of the acceleration of the

177   sensor in the direction of the crank arm corresponded to top dead centre (TDC) crank

178   position. To synchronise the EMG data with the kinematic and kinetic data, the TDC

179   locations from the accelerometer on the crank arm were matched to the corresponding TDC

180   measured by the crank encoder.

181   The raw EMG signals for the sprint efforts were high pass filtered (Butterworth second order,

182   cut off frequency 30 Hz) to diminish motion artefacts (De Luca, Gilmore, Kuznetsov, & Roy,

183   2010), root mean squared (RMS, 25 ms window) and then low pass filtered (Butterworth

184   second order, cut off frequency 24 Hz) (Brochner Nielsen et al., 2018). The data were then

185   interpolated to 100 data points around the crank cycle and then averaged over 6 crank

186   revolutions to create a linear envelope for each muscle. The EMG signals were normalised to

187   the mean value in the linear envelope across the crank cycle for each muscle.

188   *Statistical Analysis*

189   In order to test for any systematic change in performance between-sessions (for example, due

190   to learning or fatigue effects) paired *t*-tests were used to compare differences between

191   discrete values. Paired *t*-tests only test if there is a statistically significant bias between-

192   sessions (systematic change) but provide no indication of the random error due to biological

193   or mechanical variation between-sessions (Atkinson & Nevill, 1998). Similarly, differences

194   in time series data (instantaneous crank powers, crank forces, joint angles, angular velocities,

195   moments, powers and normalised EMG linear envelopes) between-sessions were assessed

196   using Statistical Parametric Mapping (SPM); paired *t*-tests were used for all variables except

197   crank forces where Hotelling's paired T^2 test was used (Pataky, 2010). Crank force consists

198   of two vector components (effective and ineffective crank force), therefore a multivariate

199   statistical test was required (Pataky, 2010). The level of statistical significance was set to $p <$

200   0.05 for all tests.

201   The reliability of the discrete variables between sessions was assessed using intra-class

202   correlation coefficient (ICC) tests. ICC's were calculated using IBM SPSS Statistics Version

203   24 (IBM UK Ltd, Portsmouth, UK), based on average measures, absolute agreement, two-

204   way mixed effects model (ICC (3,*k*) - where *k* is equal to the number of trials in a session

205   which in this study is three). The ICCs were interpreted using Koo and Li's guidelines: values

206   less than 0.50 are indicative of poor reliability, between 0.50 and 0.75 indicates moderate

207   reliability, 0.75 to 0.90 indicates good reliability and > 0.90 indicates excellent reliability

208   (Koo & Li, 2016). For a variable to be considered as having excellent reliability, both upper

209   and lower bounds of the 95% confidence intervals must fall within the excellent range (i.e. >

210   0.9) (Koo & Li, 2016).

211   Standard error of measurement (SEM) for between sessions was calculated using the formula

212   (Weir, 2005), where SD is standard deviation of the mean difference:

213    $$SEM = SD\sqrt{1 - ICC}$$

214   Minimal detectable difference (MDD) was calculated for between sessions using the formula

215   (Weir, 2005):

$$MDD = SEM \times 1.96 \times \sqrt{2}$$

216   The coefficient of variation (CV) was calculated for the average crank power over a complete

217   revolution (Hopkins, 2000).

218   The standard error of measurement (SEM) was calculated for the kinematic and kinetic time

219   series data to evaluate the reliability of these waveforms within- and between-session using

220   the methods described in Pini, Markström, & Schelin, 2019. The mean and SD SEM for a

221   complete revolution was calculated for each variable. The EMG data were visually inspected

222   for signal quality and the frequency spectrum of the raw and filtered EMG signal calculated.

223   EMG signals with a high frequency content below 20 Hz, indicates low frequency noise due

224   to movement artefact (De Luca et al., 2010) and therefore, these trials were discarded. The

225   SEM for within- and between-session for the EMG linear envelopes of the VL, VM, ST, and

226   GMAX muscles were calculated using 13 participants. At least 2 trials for each muscle per

227   session per participant were required to calculate SEM. The calculated reliability of the EMG

228   data is therefore the upper bound, as very noisy trials were discarded.

229   The cross-correlation coefficient (R) was calculated to compare the temporal effects of

230   within- and between-session EMG linear envelopes (Wren, Do, Rethlefsen, & Healy, 2006).

231   The between-sessions cross-correlation coefficient was calculated comparing the session

232   mean EMG linear envelope, and within-session the cross-correlation coefficient was

233   calculated comparing the EMG linear envelope for two trials.

234 **Results**

235 *Discrete variables*

236 Discrete crank level variables demonstrated good to excellent between-sessions reliability

237 ICC(3,*k*) > 0.756 (Table 1). Average crank power for a complete revolution for the left side

238 only was 445.3 ± 95.7 and 438.8 ± 111.5 W for session 1 and 2 respectively (Table 1), which

239 gives an indicative total power for a complete revolution, for both cranks, of 891 and 878 W.

240 MDD between-sessions for peak crank power and forces was 21 W and between 9 to 72 N

241 respectively (Table 1). Peak joint angle values typically demonstrated moderate to excellent

242 reliability, with MDD between-sessions from 1.1 to 4.4° (Table 1). Peak joint angular

243 velocity between-sessions reliability was typically moderate to excellent, except for peak

244 knee flexion and hip extension angular velocity which had poor to good reliability (Table 1).

245 MDD between-sessions for peak joint angular velocities ranged from 14 to 59°/s (Table 1).

246 Peak joint moments demonstrated moderate to excellent between-sessions reliability, except

247 for peak knee flexion moment which demonstrated poor to moderate reliability (Table 1).

248 Maximum ankle and knee joint powers demonstrated good to excellent reliability between-

249 sessions whereas, maximum hip power showed poor to good reliability (Table 1). MDD

250 between-sessions for peak joint moments ranged from 2 to 26 N.m and for maximum joint

251 powers 30 to 144 W.

252 Insert Table 1

253 CV for average crank power over a revolution was 3.0 ± 1.5% and 4.6 ± 1.9% for within- and

254 between-session respectively.

255 *Time Series Variables*

256 Crank power demonstrated excellent within- and between-session reliability, with a mean

257 SEM between-sessions over a complete revolution of 46.6 ± 9.4 W (Figure 2, Figure 3).

258 Crank power was significantly different ($p < 0.05$) between sessions one and two, between

259 crank angles 340 to 6° (7.2% of crank cycle) (Figure 2). The ineffective crank force was less

260 repeatable (mean SEM = 31.6 ± 18.2 N) than effective crank force (mean SEM = 19.8 ± 4.0

261 N) within- and between-session, which was associated with a large SEM for ineffective crank

262 force between crank cycles of 140° and 210° (Figure 4, Figure 5). The crank forces were

263 significantly different ($p < 0.05$) between sessions one and two, between crank angles 191 to

264 199° (2.2% of crank cycle), and 347 and 1° (3.9% of crank cycle) (Figure 4).

265 Joint angles and angular velocities demonstrated excellent within- and between-session

266 reliability (mean SEM $\geq 2.4°$ and 34.1°/s) (Figure 6). Ankle joint angles and angular

267 velocities were less repeatable than those at the knee and hip joints. Ankle joint angular

268 velocity was significantly different ($p < 0.05$) between sessions one and two, between crank

269 angles 152 to 170° (5.0% of crank cycle) (Figure 6).

270 Joint moments and powers demonstrated reasonable within- and between-session reliability

271 (mean SEM ≥ 15.5 N.m and 62.6 W) (Figure 6, Figure 7). Hip joint moments and powers

272 were less repeatable than those at the knee and ankle joints, particularly around the location

273 of maximum hip extension moment and power (Figure 7). Ankle joint moment was

274 significantly different ($p < 0.05$) between sessions one and two, between crank angles 340 to

275 6° (7.2% of crank cycle) (Figure 6). Hip joint power was significantly different ($p < 0.05$)

276 between session one and two between crank angles 340 to 2° (6.1% of crank cycle) (Figure

277 6).

278   EMG linear envelope normalised to the mean value in the signal demonstrated high within-

279   and between-session reliability (Figure 8). Mean SEM values for EMG linear envelopes

280   ranged between 0.14 to 0.16, and 0.16 to 0.20 proportion of the mean EMG signal, for

281   within- and between-session respectively. The GMAX, TA, and BF muscles demonstrated

282   the lowest reliability for EMG activity, and the VL and VM muscles the highest reliability

283   (Figure 8). The cross-correlation coefficient (R) which compares timing of EMG linear

284   envelopes between-sessions ranged from 0.976 to 0.990 (Figure 8).

285   Insert Figure 2, Figure 3, Figure 4, Figure 5, Figure 6, Figure 7, Figure 8

286   **Discussion and implications**

287   The purpose of this study was to quantify the test-retest reliability of kinematic, kinetic, and

288   EMG muscle activation variables measured during short-term maximal sprint cycling. Our

289   main findings were that between-sessions test-retest reliability level was typically moderate

290   to excellent for the biomechanical variables that describe maximal cycling, and furthermore

291   that within-session reliability was better than between-sessions reliability. However, some

292   variables, such as peak knee flexion moment and maximum hip joint power demonstrated

293   lower reliability, indicating that care needs to be taken when using these variables to evaluate

294   changes in maximal cycling biomechanics.

295   Within- and between-session values of SEM for joint angles and angular velocities

296   demonstrated high reliability (Figure 6). We found that ankle joint kinematics (angle and

297   angular velocity) were less repeatable than knee and hip joint kinematics, evidenced by the

298   larger mean SEM values for the ankle joint kinematics. The source of the lower reliability in

299   our ankle joint kinematics data is not clear, although it seems unlikely to be a measurement

300   error, given that anatomical landmark marker placement errors for the lower limb are greatest

301 at the hip, rather than the ankle joint (intra-examiner precision for the greater trochanter

302 marker is 12.2 mm along the long axis of the femur, and 11.1 mm in the anterior-posterior

303 direction, compared to lateral malleolus - 2.6 mm along the long axis fibula, 2.4 mm anterior-

304 posterior direction) (Della Croce, Cappozzo, & Kerrigan, 1999; Della Croce, Leardini,

305 Chiari, & Cappozzo, 2005). Furthermore, the soft tissue artefact (STA) of the lower limb

306 markers in cycling is also largest for the hip rather than the ankle joint (greater trochanter

307 marker displacement at 30 rpm submaximal cycling, 37.3 mm anterior-posterior and 10.3 mm

308 proximal-distal, compared to the lateral malleolus 15.8 mm anterior-posterior and 8.6 mm

309 proximal-distal) (Li et al., 2017). By comparison there are potential biological explanations

310 for the lower reliability of the ankle joint kinematics. Martin and Nichols, for example,

311 demonstrated that the ankle has a different role to the knee and hip joints in maximal cycling

312 and acts to transfer - instead of maximise power (Martin & Nichols, 2018). More specifically,

313 the ankle works in synergy with the hip joint to transfer power produced by the muscles

314 surrounding the hip joint to the crank (Fregly & Zajac, 1996). Our results support this notion

315 by suggesting that cyclists may regulate their ankle angle as part of this hip-ankle synergy, in

316 order to maintain a stable effective crank force. A specially designed experiment would be

317 required to test this hypothesis.

318 In terms of joint kinetics, joint moments and powers demonstrated lower reliability at more

319 proximal compared to distal joints – with the largest values of SEM for the hip joint moment

320 (Figure 6, Figure 7). This observation may be due to the STA and skin marker misplacement

321 errors being largest at the hip joint, as discussed above (Della Croce et al., 1999; Li et al.,

322 2017). It may also be due to the fact that measurement errors in general (STA, marker

323 misplacement, force pedal measurement precision) will propagate through the inverse

324 dynamics calculations (Myers, Laz, Shelburne, & Davidson, 2015). In either scenario, this

325   indicates that the observed differences in proximal to distal joint reliability are likely to be

326   due to measurement error, rather than biological variability.

327   The peak knee flexion moment showed poor to moderate between-sessions reliability, with

328   the largest MDD of all joint moments (26 N.m). Error due to knee marker misplacement is

329   dependent on knee flexion angle, with previous studies demonstrating that the greater the

330   knee flexion, the larger error in the joint angle (Della Croce et al., 1999). Marker

331   displacement could therefore explain the poor reliability of our peak knee flexion angular

332   velocity and moment data. Further work is required, using more detailed marker sets and

333   models of STA, to reduce the influence of STA and skin marker misplacement on the

334   calculated kinematics and kinetic variables, which may improve the reliability of the

335   calculated knee flexion and hip joint variables.

336   Average crank power output over a complete revolution was highly reliable both within- and

337   between-session, supporting the findings of Martin and colleagues that trained cyclists are

338   able to reproduce reliable maximal crank power within one testing session (Martin et al.,

339   2000). Effective crank force exhibited similar reliability to crank power, whereas ineffective

340   crank force demonstrated lower within- and between-session reliability which was associated

341   with the large intra-participant variability and SEM in ineffective crank force between crank

342   angles of 140° and 210° (Figure 4, Figure 5). It is unlikely that force pedals' measurement

343   precision would provide an explanation for these observed differences in reliability between

344   the effective and ineffective crank forces, given that the measurement precision values are the

345   same for all components of force for the instrumented pedals we used (combined error -

346   linearity and hysteresis 1% measuring range (MR) and crosstalk between the components

347   (<1.5% MR) (Sensix, Poitiers, France)). Therefore, it seems probable that the reliability

348   difference between effective and ineffective force may have a biological basis, a notion

349   which can be expanded upon using our EMG results.

350   EMG linear envelopes generally demonstrated excellent reliability (Figure 8). However, the

351   GMAX, BF and the TA muscles demonstrated the lowest reliability for EMG activity. Lower

352   reliability of the EMG activity for the GMAX and TA muscles have been demonstrated in

353   submaximal cycling (Jobson et al., 2013). The between-sessions reliability of the EMG

354   activity of the GMAX muscle has been shown to decrease with increasing workload

355   (between-sessions CV = 43.1% at 265 W compared to CV = 23.0 at 135 W) (Jobson et al.,

356   2013) which might suggest greater biological variation in the GMAX muscle activity with

357   increased workload, potentially explaining the lower reliability of the GMAX EMG activity.

358   Jobson and colleagues suggested the lower reliability of the EMG activity for the TA muscle

359   might be owing to the fact some cyclists have two bursts of muscle activity per crank

360   revolution which may introduce more between crank revolution variability (Jobson et al.,

361   2013). Measurement error could also be a potential source of the lower reliability of the EMG

362   activity for the TA, as the location of the EMG sensor can strongly influence the pattern of

363   EMG activity recorded owing to crosstalk from the peroneus longus muscle during dynamic

364   movements (Campanini et al., 2007; Hug, 2011). Therefore, small changes in positioning of

365   the EMG sensor between sessions could influence the EMG activity measured. Wren and

366   colleagues suggested the lower reliability of the hamstrings may be due to measurement error

367   reflecting the increased sensitivity of these muscles to electrode placement owing to muscle

368   length and overlying fat mass (Wren et al., 2006). The lower reliability of EMG activity in

369   the BF hamstring muscle may also have a biological basis however, given that our findings

370   are consistent with other studies who suggest that this is related to their bi-articular function

371   (Ryan & Gregor, 1992). Van Ingen Schenau and colleagues for example demonstrated that

372 the bi-articular muscles are important for controlling the direction of the external force on the

373 pedal (van Ingen Schenau, Boots, De Groot, Snackers, & Van Woensel, 1992). They

374 identified that the paradoxical coactivation of the mono-articular agonists (vastii) with bi-

375 articular antagonists (hamstrings) emerges so the bi-articular muscles can help control the

376 desired direction of the force applied to the pedal by adjusting the relative distribution of net

377 moments over the joints (van Ingen Schenau et al., 1992).

378 On a mechanical basis, the goal of maximal cycling is to maximise the effective crank force

379 as this maximises the propulsive power and thus the speed of the bicycle. Taking our crank

380 force and EMG data together therefore, our results allow us to speculate that cyclists may

381 regulate bi-articular muscles activation to control the direction of the pedal force, with the

382 aim of maximising effective crank force and maintaining a stable outcome at the expense of

383 the ineffective force which does not directly affect the task outcome. The bi-articular muscles

384 (BF, ST and GL) are active in the region of the crank cycle where the ineffective crank is

385 more variable which could explain the biological mechanism underlying this finding. This

386 principle has been observed in walking (Kadaba et al., 1989; Giakas & Baltzopoulos, 1997)

387 and running (Kinoshita, Bates, & DeVita, 1985), where the propulsion and braking ground

388 reaction forces (anterior-posterior and vertical direction) have been shown to have lower

389 between-stride variability than the medio-lateral force. However, further, purposefully

390 designed experiments are required to confirm or refute these speculations.

391 SPM indicated a significant between-session difference for small regions of the crank cycle,

392 for crank power, crank forces, ankle angular velocity and moment, and hip power. These

393 differences are unlikely to be meaningful changes as these are less than 7.2% of the crank

394 cycle, and typically occur in regions of low magnitude in these variables.

395   The experimental protocol could have introduced some variability to the kinematics, as

396   although the participants were instructed to remain seated during the sprints on the ergometer,

397   they tended to hover slightly over the saddle (potentially with the aim to increase crank

398   power), which increases pelvis movement. Also, the ergometer was set-up to match each

399   participant's track bike. Therefore, saddle height was not standardised to percentage of inside

400   leg length, which is often recommended (de Vey Mestdagh, 1998). Some of the participants

401   had a relatively low saddle height compared to their leg length, which resulted in relatively

402   large pelvis obliquity (rocking) and transverse rotation when they sprinted. This strategy may

403   have introduced more within- and between-trial variability, particularly at the hip joint. We

404   acknowledge that we measured 2D kinematics using a high-speed video camera, which is not

405   considered the 'gold standard' for measuring kinematics which is 3D motion capture systems

406   (Fonda, Sarabon, & Li, 2014). However, these methods were utilised because during cycling

407   the movement is predominantly in the sagittal plane (Umberger & Martin, 2001; van Ingen

408   Schenau, Van Woensel, Boots, Snackers, & De Groot, 1990) and therefore previous studies

409   that have investigated maximal cycling have just considered the sagittal plane actions, as this

410   is the plane where muscles produce power to generate effective crank force (Barratt, Korff,

411   Elmer, & Martin, 2011; Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014).

412   Therefore, we measured 2D kinematics in the sagittal plane using a simple marker set which

413   has the added benefit of reducing time required for data collection sessions which is an

414   important ethical consideration when working with elite athletes.

415   **Conclusion**

416   Typically, the biomechanical variables that describe maximal cycling are reliable. However,

417   some variables have lower reliability indicating that care needs to be taken when using these

418   variables to evaluate changes in maximal cycling biomechanics. Our results allow us to

419 speculate that biological variability is the source of the lower reliability of the ineffective

420 crank force, ankle kinematics and hamstring muscles activation while measurement error is

421 the source of the lower reliability in hip and knee joint kinetics. Further research using

422 purposefully designed experiments is required to confirm or refute these speculations. We

423 recognise that there were some data collection problems (noisy EMG data and no right force

424 pedal data) which might indicate potentially lower reliability of our data collection method.

425 These reliability data can be used to help understand the practical relevance of a longitudinal

426 intervention on athletes' maximal cycling performance.

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577

Table 1: Between-sessions reliability for kinematic and kinetic variables, * indicates significant difference between sessions ($p < 0.05$), ICC(3,k) = Between-sessions intraclass correlation with lower (LB) and upper (UB) bound confidence intervals, SEM = standard error of measurement, MDD = minimal detectable difference

Variable	Units	Mean(SD)	Mean difference	p	ICC	95%	95%	SEM	MDD

|		Session 1	Session 2			(3,k)	LB	UB		
Power (average for left crank)	W	445.3 ± 95.7	438.8 ± 111.5	-6.5	0.429	0.979	0.938	0.993	4.3	12
Pedalling rate	rpm	134.8 ± 1.3	134.7 ± 1.4	-0.2	0.021*	0.986	0.935	0.996	0.0	0.1
Max effective crank force	N	593.3 ±126.2	579.0 ± 130.9	-14.4	0.072	0.986	0.952	0.996	3.2	9
Max ineffective crank force	N	603.5 ± 172.1	605.3 ± 165.4	1.8	0.944	0.923	0.756	0.975	25.9	72
Min ineffective crank force	N	-192.7 ± 65.2	-207.3 ± 82.3	-14.7	0.136	0.937	0.805	0.980	8.7	24
Max instantaneous crank power	W	1387.2 ± 309.2	1348.4 ± 316.5	-38.7	0.043*	0.986	0.946	0.996	7.7	21
Peak ankle plantarflexion angle	°	141.7 ± 11.3	142.3 ± 11.5	0.6	0.446	0.983	0.948	0.994	0.4	1.1
Peak ankle dorsiflexion angle	°	113.1 ± 5.0	113.8 ± 5.8	0.7	0.281	0.955	0.863	0.985	0.5	1.3
Peak knee extension angle	°	142.7 ± 6.4	143.5 ± 5.7	0.8	0.489	0.864	0.580	0.956	1.6	4.4
Peak knee flexion angle	°	70.0 ± 3.6	70.2 ± 3.4	0.2	0.715	0.857	0.550	0.954	1.0	2.6
Peak hip extension angle	°	68.1 ± 5.0	68.4 ± 4.6	0.3	0.720	0.893	0.665	0.966	1.0	2.8
Peak hip flexion angle	°	26.1 ± 4.3	25.6 ± 4.2	-0.5	0.447	0.916	0.746	0.973	0.7	1.9
Peak ankle plantarflexion angular velocity	°/s	236.6 ± 65.7	247.1 ± 65.0	10.4	0.441	0.839	0.509	0.948	19.7	55
Peak ankle dorsiflexion angular velocity	°/s	-262.0 ± 91.2	-268.5 ± 107.2	-6.6	0.561	0.957	0.868	0.986	8.6	24
Peak knee extension angular velocity	°/s	472.8 ± 43.2	479.1 ± 33.8	6.3	0.434	0.838	0.504	0.948	11.8	33
Peak knee flexion angular velocity	°/s	-507.5 ± 57.6	-513.3 ± 43.6	-5.8	0.635	0.772	0.279	0.927	21.4	59
Peak hip extension angular velocity	°/s	265.6 ± 29.1	273.8 ± 21.9	8.2	0.141	0.814	0.447	0.939	8.5	24
Peak hip flexion angular velocity	°/s	-277.6 ± 30.7	-273.4 ± 35.1	4.2	0.390	0.924	0.769	0.975	4.9	14
Peak ankle plantarflexion moment	N.m	78.6 ± 18.6	81.4 ± 20.2	2.8	0.372	0.910	0.729	0.971	3.4	9
Peak ankle dorsiflexion moment	N.m	-14.0 ± 7.0	-12.3 ± 6.0	1.8	0.049*	0.928	0.743	0.978	0.8	2
Peak knee extension moment	N.m	90.0 ± 34.5	82.9 ± 33.5	-7.1	0.028*	0.965	0.852	0.990	2.0	6
Peak knee flexion moment	N.m	-50.7 ± 20.9	-57.7 ± 15.0	-7.0	0.151	0.697	0.127	0.900	9.4	26
Peak hip extension moment	N.m	132.3 ± 30.7	140.4 ± 32.8	8.1	0.086	0.919	0.737	0.974	4.6	13
Peak hip flexion moment	N.m	-47.7 ± 26.1	-41.3 ± 17.0	6.5	0.115	0.870	0.600	0.958	5.1	14
Maximum ankle power	W	259.6 ± 111.7	258.5 ± 107.8	-1.1	0.937	0.951	0.846	0.984	10.9	30
Maximum knee power	W	659.6 ± 321.7	620.4 ± 253.6	-39.2	0.160	0.968	0.901	0.990	17.6	49
Maximum hip power	W	519.8 ± 186.3	578.1 ± 153.0	58.3	0.104	0.826	0.474	0.944	52.1	144

Figure captions
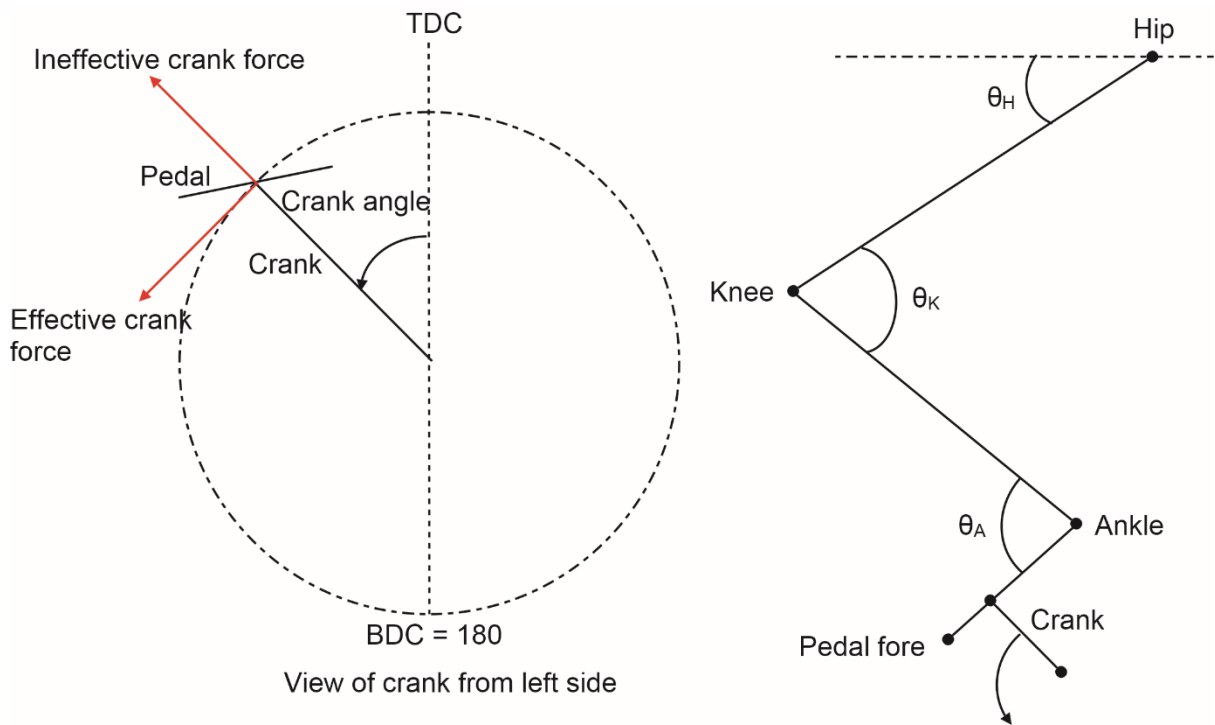
Figure 1: Joint angle and crank forces convention. TDC = top dead centre, BDC = bottom dead centre, θ_H = hip angle, θ_K = knee angle, θ_A = ankle angle
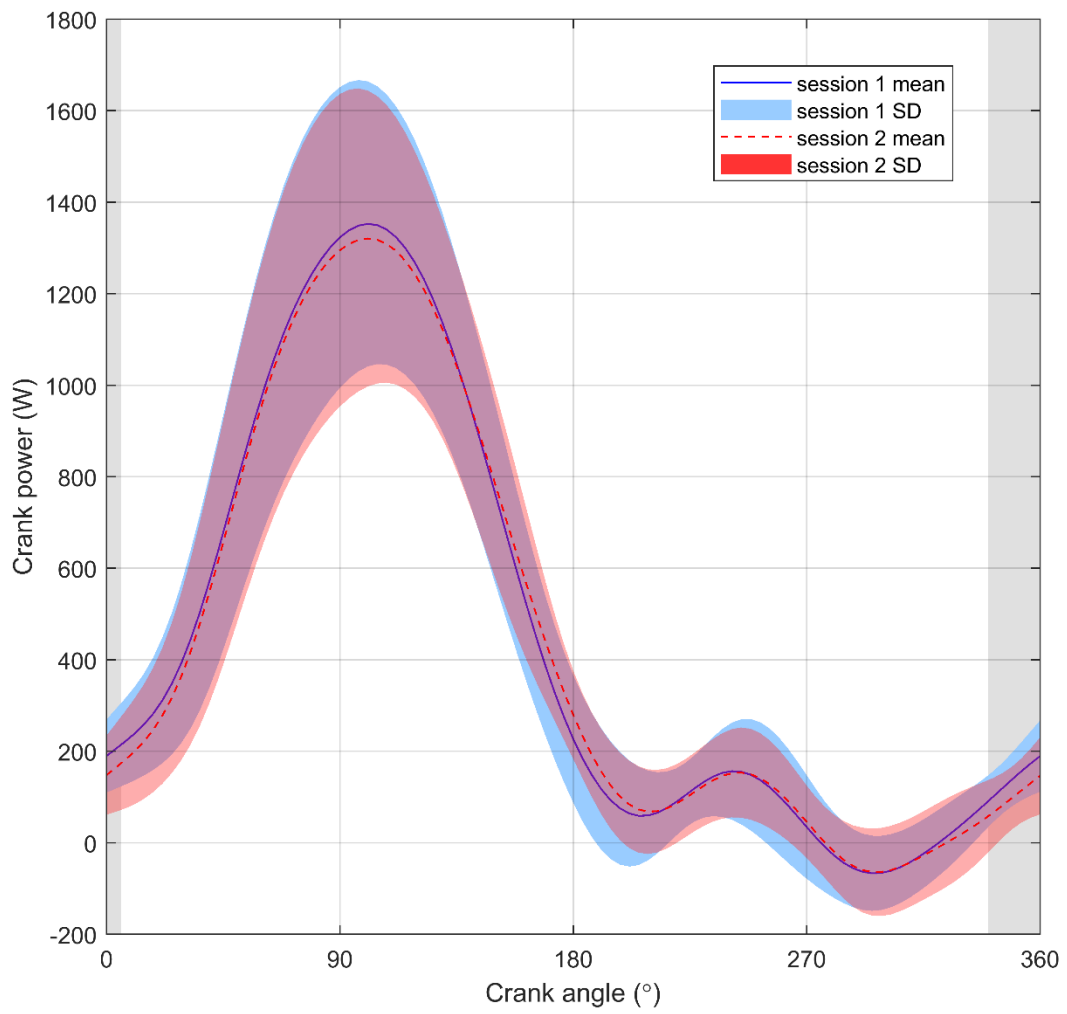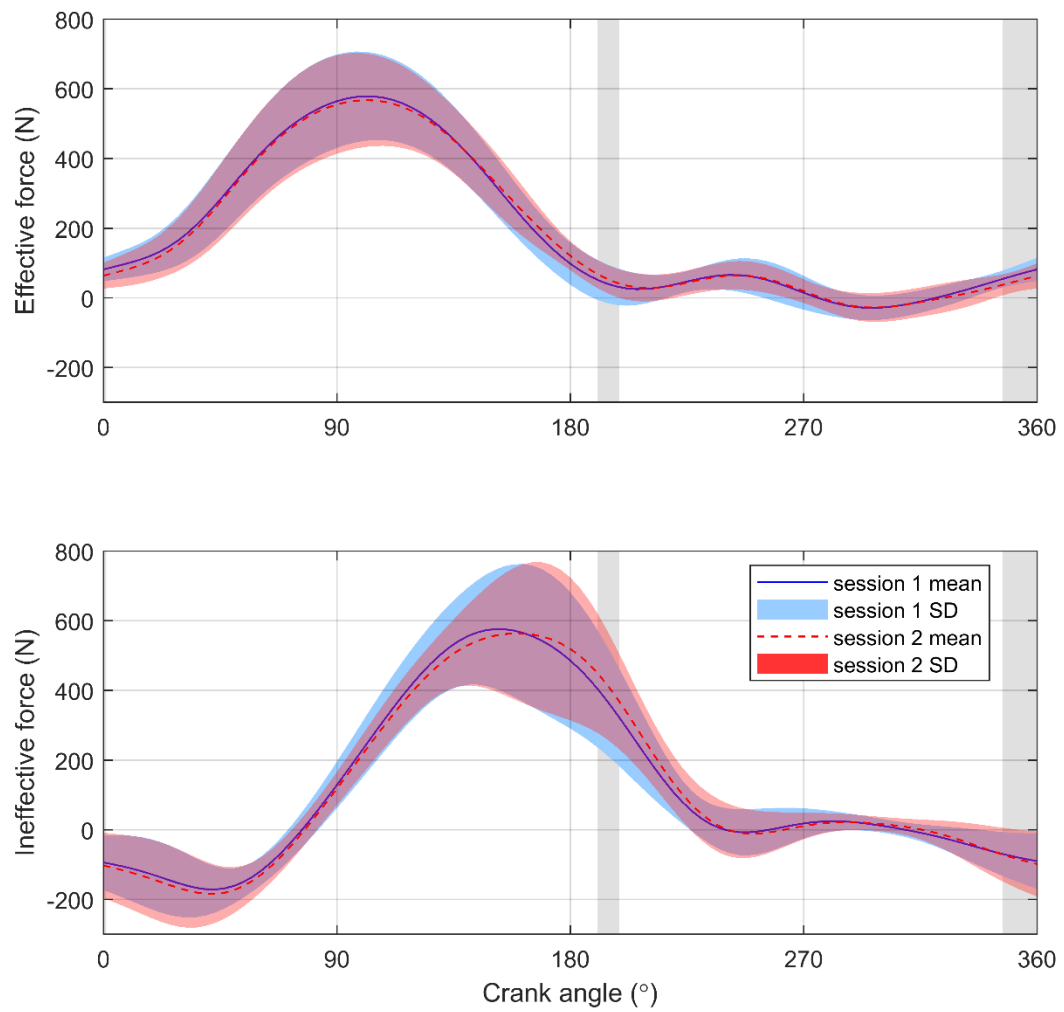
Figure 2: Crank power: group means for session one and two. Areas of the graph shaded grey where the Statistical parametric mapping (SPM) is significant.
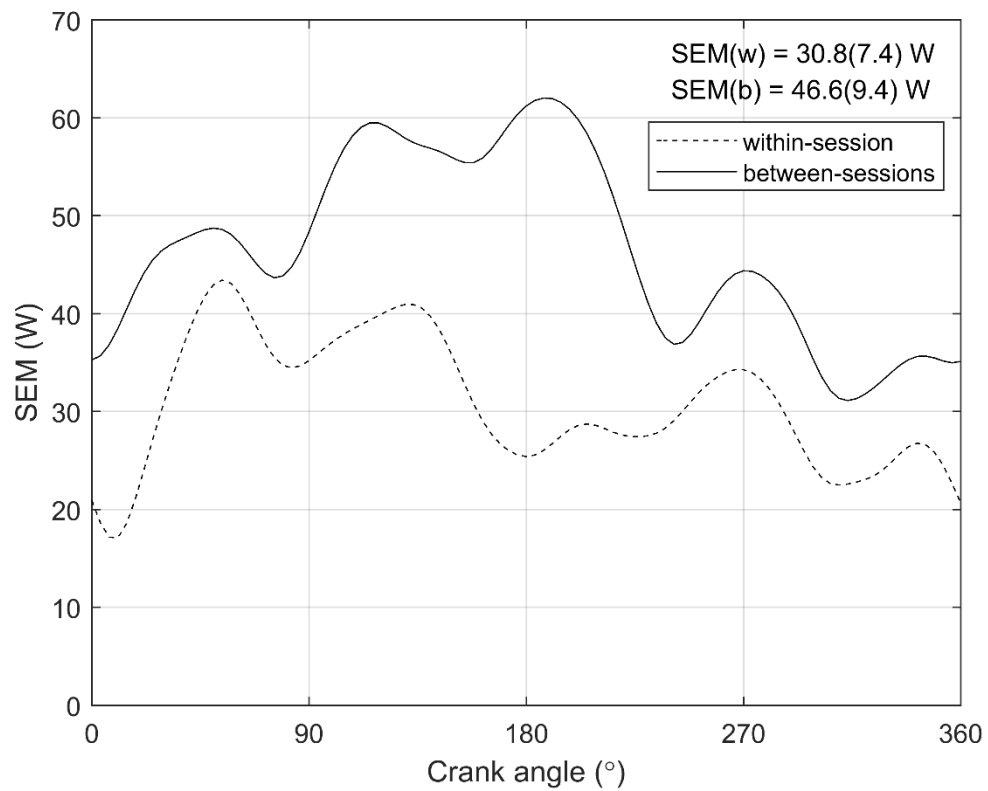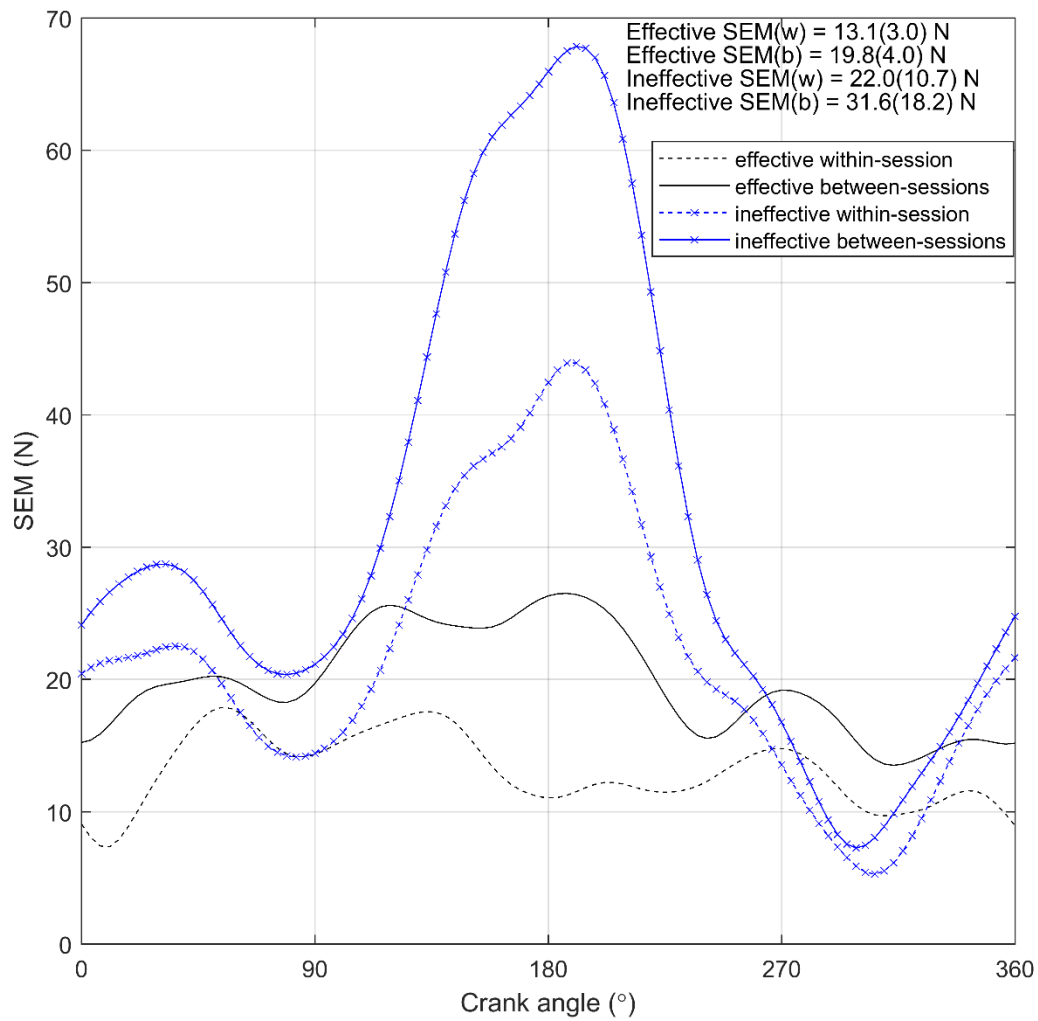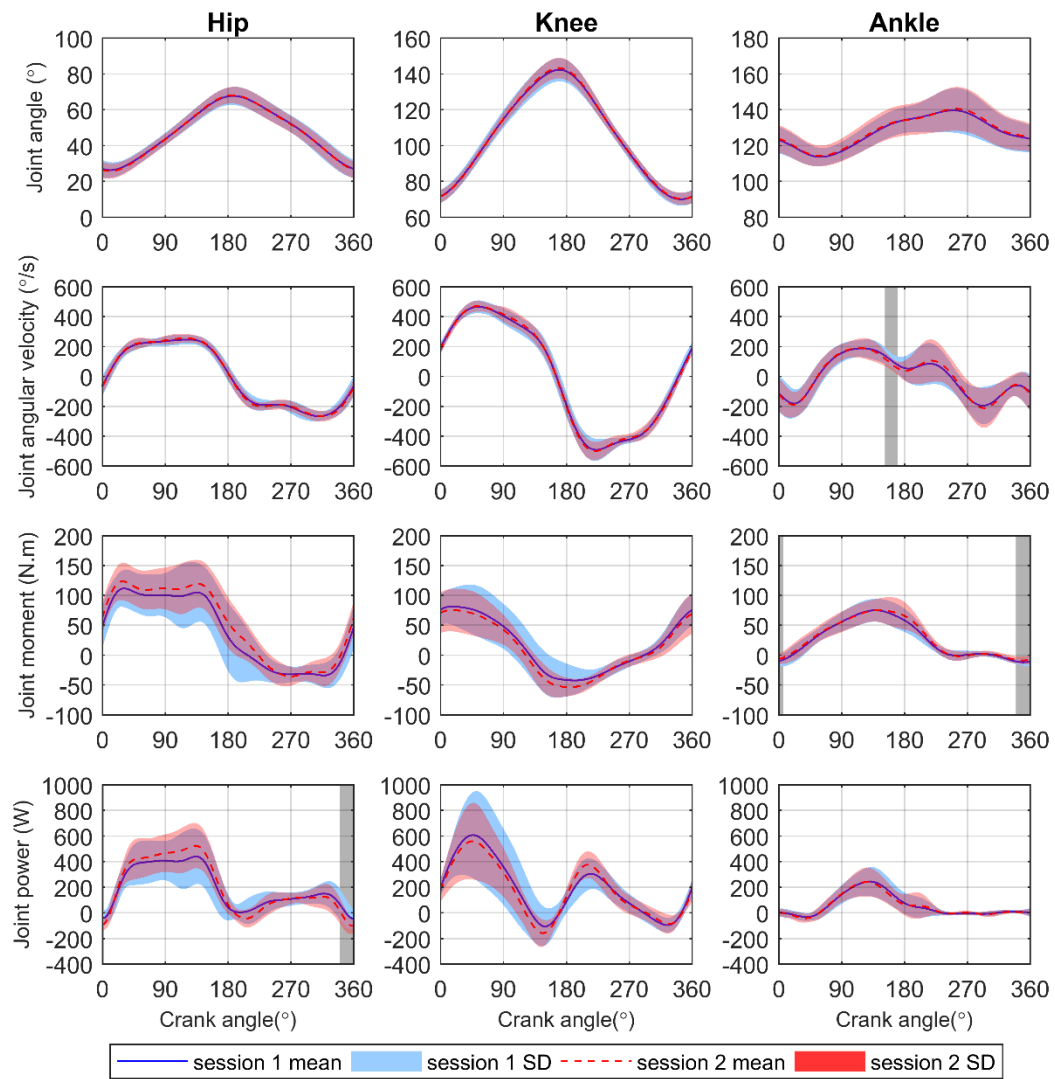
Figure 3: Crank power: standard error of measurement (SEM) within- and between-session. Mean and standard deviation of SEM within-session (w) and between-sessions (b) over complete crank cycle.

Figure 4: Crank forces: group means for session one and two. Areas of the graph shaded grey where the Statistical parametric mapping (SPM) is significant.
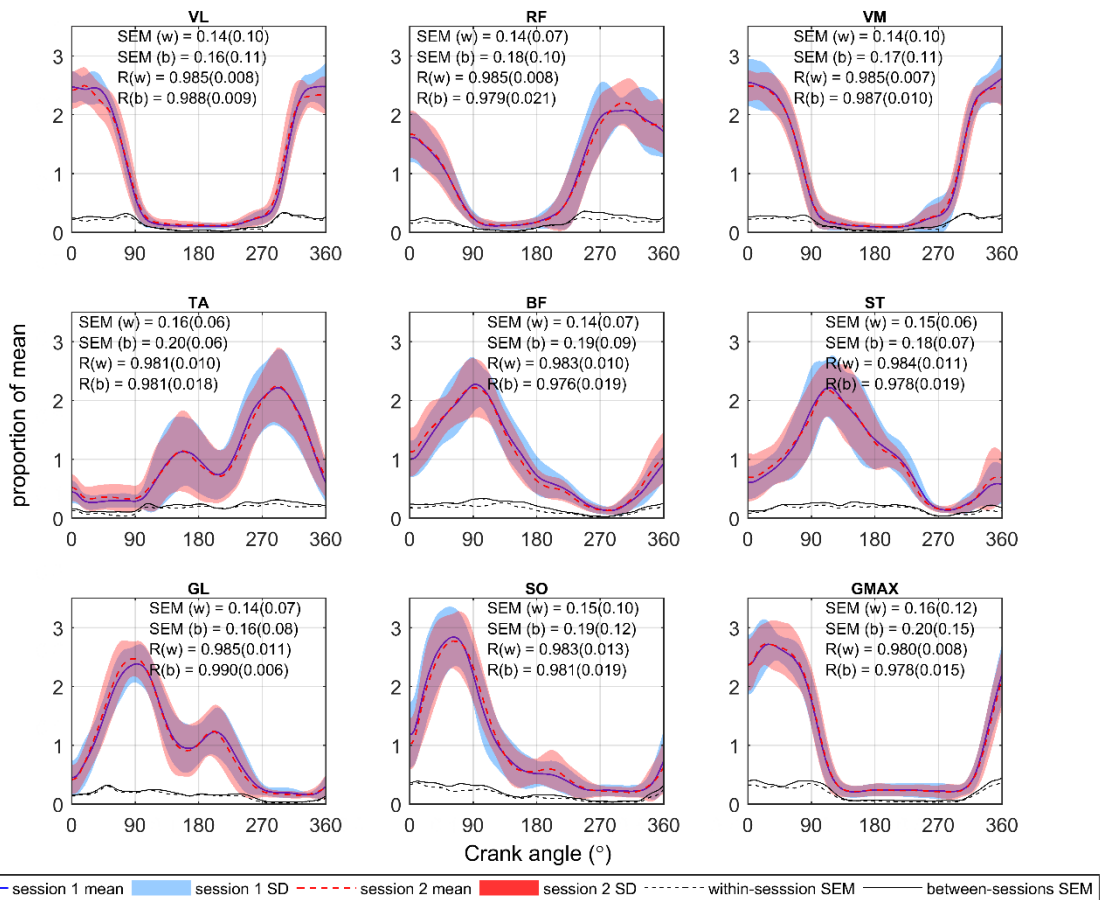
Figure 5: Crank forces: standard error of measurement (SEM) within- and between-session. Mean and standard deviation of SEM within-session (w) and between-sessions (b) over complete crank cycle.

Figure 6: Joint angles, angular velocities, moments and powers: group means for session one and two. Areas of the graph shaded grey where the Statistical parametric mapping (SPM) is significant.
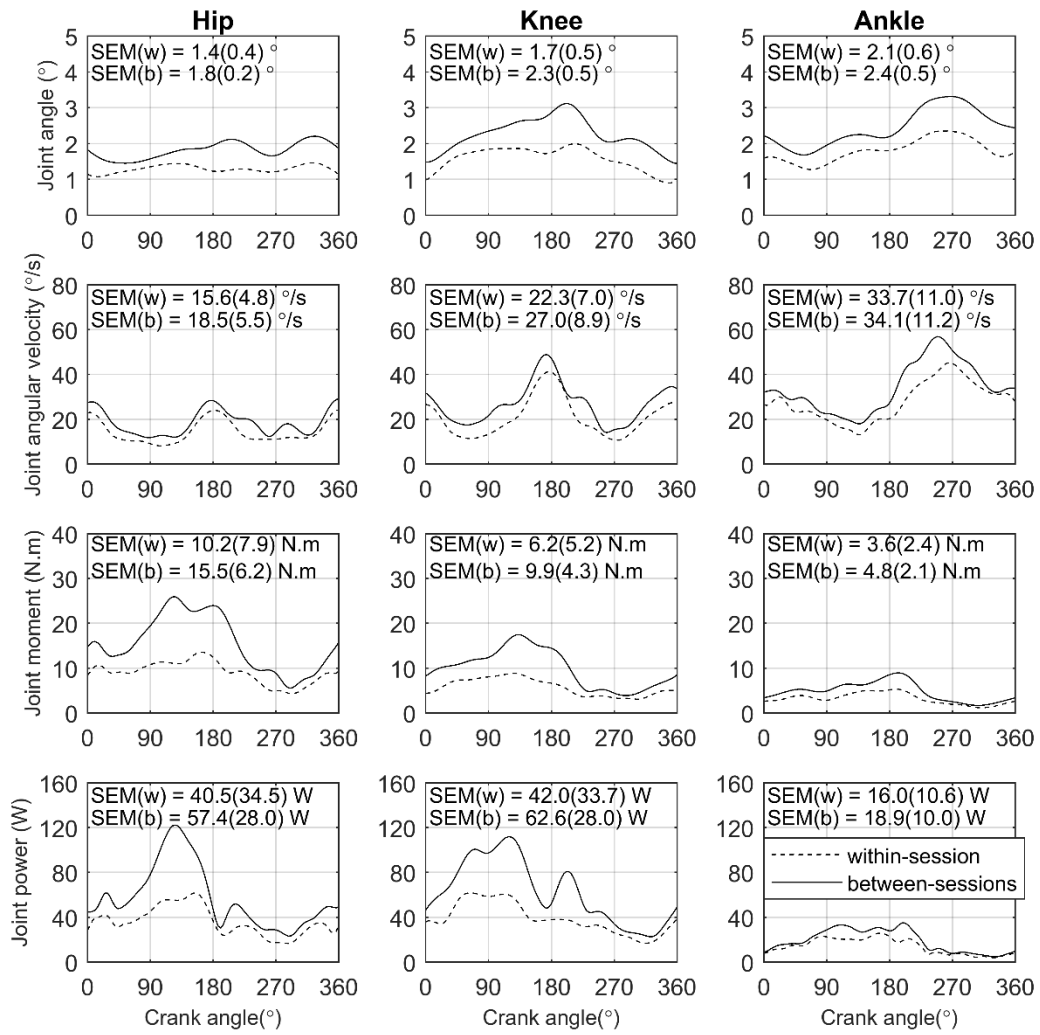
Figure 7: Joint angles, angular velocities, moments and powers: standard error of measurement (SEM) within- and between-session. Mean and standard deviation of SEM within-session (w) and between-sessions (b) over complete crank cycle.

Figure 8: EMG linear envelopes (normalised to mean value in signal) for each muscle: group means for session one and two and standard error of measurement (SEM) within- and between-session. VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF=biceps femoris, ST= semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Mean and standard deviation of SEM within-session (w) and between-sessions (b) over complete crank cycle.