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Title:

Musculoskeletal model-based inverse dynamic analysis under ambulatory conditions using inertial motion capture

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1 **Abstract**

2 Inverse dynamic analysis using musculoskeletal modeling is a powerful tool,
3 which is utilized in a range of applications to estimate forces in ligaments, mus-
4 cles, and joints, non-invasively. To date, the conventional input used in this
5 analysis is derived from optical motion capture (OMC) and force plate (FP)
6 systems, which restrict the application of musculoskeletal models to gait labo-
7 ratories. To address this problem, we propose the use of inertial motion cap-
8 ture to perform musculoskeletal model-based inverse dynamics by utilizing a
9 universally applicable ground reaction force and moment (GRF&M) prediction
10 method. Validation against a conventional laboratory-based method showed
11 excellent Pearson correlations for sagittal plane joint angles of ankle, knee, and
12 hip ($\rho = 0.95, 0.99, \text{ and } 0.99$, respectively) and root-mean-squared-differences
13 (RMSD) of $4.1 \pm 1.3^\circ$, $4.4 \pm 2.0^\circ$, and $5.7 \pm 2.1^\circ$, respectively. The GRF&M pre-
14 dicted using IMC input were found to have excellent correlations for three com-
15 ponents (vertical: $\rho = 0.97$, $\text{RMSD} = 9.3 \pm 3.0 \text{ \%BW}$, anteroposterior: $\rho = 0.91$,
16 $\text{RMSD} = 5.5 \pm 1.2 \text{ \%BW}$, sagittal: $\rho = 0.91$, $\text{RMSD} = 1.6 \pm 0.6 \text{ \%BW} \cdot \text{BH}$), and
17 strong correlations for mediolateral ($\rho = 0.80$, $\text{RMSD} = 2.1 \pm 0.6 \text{ \%BW}$) and
18 transverse ($\rho = 0.82$, $\text{RMSD} = 0.2 \pm 0.1 \text{ \%BW} \cdot \text{BH}$). The proposed IMC-based
19 method removes the complexity and space-restrictions of OMC and FP systems
20 and could enable applications of musculoskeletal models in either monitoring
21 patients during their daily lives or in wider clinical practice.

22 1. Introduction

23 Assessment of muscle, joint, and ligament forces is important to understand
24 the mechanical and physiological mechanisms of human movement. To date,
25 the measurement of such in-vivo forces is a challenging task. For this reason,
26 computer-based musculoskeletal models have been widely used to estimate the
27 variables of interest non-invasively [1, 2].

28 The most common approach used in musculoskeletal modeling is the method
29 of the inverse dynamics [3]. This analysis utilizes the equations of motion with
30 input from human body kinematics in conjunction with kinetics obtained from
31 external forces [4], to estimate joint reaction and muscle forces, as well as net
32 joint moments using muscle recruitment methods [5]. Measurements of the
33 external forces are typically required and measured using force plates (FPs),
34 however, the use of FPs has several limitations. First, subjects tend to alter
35 their natural gait patterns in order to hit the small and fixed measurement area
36 of a plate [6]. In addition, this static and limited measurement area, impedes
37 the assessment of several consecutive steps, when only a couple of FPs are
38 available. Finally, the combined use of FP with motion input introduces a
39 dynamic inconsistency, which results to residual forces and moments in the
40 inverse dynamics. [7, 8].

41 Several studies have proposed replacing the FP input with wearable de-
42 vices such as shoes with three-dimensional force and torque sensors beneath
43 the sole [9, 10, 11]. In a similar fashion, pressure insoles were proposed to re-
44 construct the complete ground reaction forces and moments (GRF&M) from
45 pressure distributions [12, 13, 14]. Although these wearable devices are suitable
46 for the assessment of external forces, the increased height and weight of the
47 shoes equipped with force/torque sensors [15, 16], as well as the repeatability
48 of the pressure sensors [17] are considered important limitations.

49 Recent research has suggested the replacement of the force input with predic-
50 tions derived solely from motion input [18, 19, 20, 21, 22, 23]. In these studies,
51 human body kinematics are combined with the inertial properties of the body

52 segments, from which Newton-Euler equations are utilized to compute the exter-
53 nal forces and moments. Since the system of equations becomes indeterminate
54 during the double stance of gait, each of the aforementioned studies focused on
55 methods to solve this issue. Ren *et al.* [19] suggested a gait event-based func-
56 tion which is only applicable in gait, while Oh *et al.* [20] and Choi *et al.* [21]
57 suggested methods based on a machine learning that require a training database
58 and thus are not applicable for movements not included in that database. A
59 last approach enables the universal application of these methods using a muscle
60 recruitment approach has shown promising performance for various activities of
61 daily living [22] and sports [23].

62 The majority of the existing research which studied the prediction of GRF&M,
63 used conventional optical motion capture (OMC) input. Despite the high ac-
64 curacy of this method in tracking marker trajectories, its dependence on lab-
65 oratory equipment restricts possible applications during daily life activities or
66 in wider clinical practice. In the previous decade, ambulatory motion tracking
67 systems based on inertial measurement units (IMUs), have been proposed as a
68 suitable alternative for estimating 3D segment kinematics [24, 25, 26, 27]. A
69 key benefit of such systems is that they can be applied in virtually any environ-
70 ment without depending on external infrastructure, such as cameras. Driven
71 by these advances in inertial motion capture (IMC), recent work of the authors
72 demonstrated its ability to estimate three-dimensional GRF&M [28], which were
73 distributed between the feet using a smooth transition assumption concept [19].
74 However, limitations of that approach is that it is only valid for gait and has no
75 muscle, bone or ligament force estimate capabilities.

76 To date, the use of detailed musculoskeletal modeling with kinematic inputs
77 from IMUs has only received limited attention. Koning *et al.* [29] previously
78 demonstrated the feasibility of kinematically driving a musculoskeletal model
79 using only orientations from IMUs. However, that study only compared the
80 kinematics of the musculoskeletal model, without any inverse dynamic calcula-
81 tions.

82 The aim of this study was to develop a workflow to perform musculoskeletal

83 model-based inverse dynamics using exclusively IMC input, applicable in am-
84 bulatory environments and validate it against a conventional laboratory-based
85 approach.

86 **2. Methods**

87 *2.1. Subjects*

88 The experimental data was collected at the Human Performance Labora-
89 tory, at the Department of Health Science and Technology, Aalborg University,
90 Aalborg, Denmark following the ethical guidelines of The Scientific Ethical Com-
91 mittee for the Region of North Jutland (Den Videnskabetiske Komit for Region
92 Nordjylland). Eleven healthy male individuals with no present musculoskeletal
93 or neuromuscular disorders volunteered for the study (age: 31.0 ± 7.2 years;
94 height: 1.81 ± 0.06 m; weight: 77.3 ± 9.2 kg; body mass index (BMI): $23.6 \pm$
95 2.4 kg/m²). All participants provided written informed consent, prior to data
96 collection.

97 *2.2. Instrumentation*

98 Full-body IMC data were collected using the Xsens MVN Link (Xsens Tech-
99 nologies B.V., Enschede, the Netherlands), in which 17 IMUs were mounted on
100 the head, sternum, pelvis, upper legs, lower legs, feet, shoulders, upper arms,
101 forearms and hands using the dedicated clothing. The exact location of each
102 sensor on the respective segment followed the manufacturer guidelines described
103 in the manual of Xsens MVN [30]. The affiliated software Xsens MVN Studio
104 4.2.4 was used to track the IMU orientations with respect to an earth-based
105 coordinate frame [24, 25]. Segment orientations were obtained by applying the
106 IMU-to-segment alignment, found using a known upright pose (N-pose) per-
107 formed by the subject at a known moment in time, while taking specific care
108 to minimize the effect of magnetic disturbances. In addition, this information
109 is fused with updates regarding the joints and external contacts to limit the
110 position drift [26].

111 For validation purposes, an OMC system utilizing 8 infrared high speed
112 cameras (Oqus 300 series, Qualisys AB, Gothenburg, Sweden) and the software
113 Qualisys Track Manager 2.12 (QTM) were used to track the trajectories of 53
114 reflective markers mounted on the human body, as described in the Appendix of
115 [28]. In addition, three FP systems (AMTI OR6-7-1000, Advanced Mechanical
116 Technology, Inc., Watertown, MA, USA) embedded in the floor of the laboratory,
117 were utilized using QTM to record the GRF&Ms. Both IMC and OMC systems
118 sampled data at a frequency of 240 Hz, while the FP system sampled data at
119 2400 Hz and subsequently downsampled to 240 Hz to match the IMC and OMC
120 sampling rate.. A second-order forward-backward low-pass Butterworth filter
121 was applied to the reflective marker trajectories and measured GRF&M, with
122 cut-off frequencies of 6 Hz and 15 Hz, respectively.

123 *2.3. Experimental protocol*

124 For each participant, the body dimensions were extracted using a conven-
125 tional tape following the guidelines of Xsens. During the data collection, the
126 subjects were instructed to walk barefoot in three different walking speeds (com-
127 fortable; CW, fast; FW, and slow; SW). The walking speeds performed experi-
128 mentally were quantified as 1.28 ± 0.14 m/s (mean \pm standard deviation) for CW,
129 1.58 ± 0.09 m/s for FW (CW + 23%) and 0.86 ± 0.11 m/s for SW (CW - 33%).
130 For every walking speed, five successful trials were assessed. A successful trial
131 was obtained when a single foot hit one of the FPs entirely, followed by an entire
132 hit of the other foot on the successive FP.

133 *2.4. Overall description of the components in the musculoskeletal models*

134 Three musculoskeletal models have been constructed in AnyBodyTM Mod-
135 eling System (AMS) v.6.0.7 (AnyBodyTM Technology A/S, Aalborg, Denmark)
136 [1]:

- 137 • a model in which the kinematics are driven by IMC and the GRF&M are
138 predicted from the kinematics (IMC-PGRF).

- 139 • a model in which the kinematics are driven by OMC and the GRF&M are
140 predicted from the kinematics (OMC-PGRF).
- 141 • a model in which the kinematics are driven by OMC and the GRF&M are
142 measured from FPs (OMC-MGRF).

143 In the IMC-PGRF model, a Biovision Hierarchy (BVH) file is exported from
144 Xsens MVN Studio and imported in AMS, in which a stick figure model is ini-
145 tially reconstructed. The BVH file contains a hierarchy part with a description
146 of the linked segment model in a static pose, as well as a motion part that
147 contains, for each time frame, the absolute position and orientation of the root
148 pelvis segment, and the joint angles between the segments described in the hier-
149 archy. The generated stick figure model contains 72 degrees-of-freedom (DOF).
150 In order to match the stick figure model with the musculoskeletal model, we
151 utilize a concept of virtual markers (VMs) demonstrated in a previous Kinect-
152 based study [31]. The VMs are mapped to particular points of each model that
153 are well defined in both models, such as joint centers and segment end points.
154 The VM placement is illustrated in Figure 1 and described in more detail in
155 the supplementary material. Following this step, the VMs are treated as actual
156 experimental markers, as if they were derived from an OMC system and they
157 are assigned weights in three directions in the segmentframe. Contrary to OMC,
158 no filtering was applied to the VM trajectories.

159 In all models, the GaitFullBody template of the AnyBodyTM Managed
160 Model Repository (AMMR) 1.6.2 was used to reconstruct the musculoskele-
161 tal models in AMS. The lumbar spine model was derived from the study of
162 de Zee *et al.* [32], the lower limb model was derived from the Twente Lower
163 Extremity Model Klein-Horsman *et al.* [33], and the shoulder and upper limb
164 models were based on the model of the Delft Shoulder Group [34, 35, 36]. The
165 full-body kinematic model contained 39 DOF in total. Specifically, a pelvis
166 segment with three rotational and three translational DOF, two spherical hip
167 joints, two revolute knee joints, two universal ankle joints, a spherical pelvic-
168 lumbar joint, two glenohumeral joints with five DOF each, two universal elbow

169 joints, and two universal wrist joints. The motion of the neck joint was locked
170 to a neutral position.

171 2.5. Scaling and kinematics analysis of the musculoskeletal models

172 For each subject, a standing reference trial with an anatomical pose was
173 utilized to identify the parameters of segment lengths and the (virtual) marker
174 positions, using a least-square minimization between the model and input (vir-
175 tual or skin-mounted) marker positions [37]. In the IMC-PGRF musculoskeletal
176 model, the lengths of the shanks, thighs, head, upper arm and forearms were
177 derived directly from the stick figure, as generated from Xsens MVN studio us-
178 ing the measured body dimensions. In contrast, the pelvis width, foot length,
179 and trunk height were optimized based on the above-mentioned least-square
180 minimization method. The estimated segment lengths were used in all subse-
181 quent dynamic trials to perform the kinematic analysis based on the method of
182 Andersen *et al.* [38].

183 2.6. Inertial and geometric scaling of the musculoskeletal models

184 The mass of each segment was linearly scaled based on the total body mass
185 and the segment mass ratio values reported by Winter [4]. The inertial pa-
186 rameters were calculated by considering the segments as cylinders with uniform
187 density. In addition, geometric scaling of each segment, where the longitudinal
188 axis was defined as the second entry, was achieved using the following matrix:

$$S = \begin{bmatrix} \sqrt{\frac{m_s}{l_s}} & 0 & 0 \\ 0 & l_s & 0 \\ 0 & 0 & \sqrt{\frac{m_s}{l_s}} \end{bmatrix} \quad (1)$$

189 where S is the scaling matrix, l_s is the ratio between the unscaled and scaled
190 lengths of the segment, m_s is the mass ratio of the segment.

191 *2.7. Muscle recruitment*

192 The muscle recruitment problem was solved by defining an optimization
193 problem where a system of equations minimizes the cost function, subject to
194 the dynamic equilibrium equations and non-negativity constraints, so that each
195 muscle can only pull, but not push, while its force remains below its strength
196 [1, 31, 39].

197 The strengths of the muscles were derived from previous studies which de-
198 scribed the models of the body parts, and were considered constant for different
199 lengths and contraction velocities [32, 33, 34, 35, 36]. To scale the muscle
200 strengths, fat percentage was used as in Veeger *et al.* [35], calculated from the
201 body mass index [40]. The model of the lower body contained 110 muscles,
202 distributed into 318 individual muscle paths. In contrast, in the upper body
203 model, ideal joint torque generators were utilized. Actuators for residual forces
204 and moments with capacity up to 10 N and Nm, respectively, were placed at the
205 origin of the pelvis and included in the muscle recruitment problem previously
206 described.

207 *2.8. Ground reaction force and moment prediction*

208 The GRF&M were predicted by adjusting a method of Skals *et al.* [23]. A
209 set of eighteen dynamic contact points were overlaid 1 mm beneath the inferior
210 surface of each foot. Each dynamic contact point consisted of five unilateral
211 force actuators, which could generate a positive vertical force perpendicular to
212 the ground, and static friction forces in the anterior, posterior, medial, and
213 lateral directions using a friction coefficient of 0.5. In addition, the height and
214 velocity activation thresholds were set to 0.03 m and 1.2 m/s, respectively.

215 *2.9. Data Analysis*

216 Lower limb joint angles calculated in the IMC-PGRF model were compared
217 to the OMC-PGRF/OMC-MGRF. In addition, GRF&M and JRF&M of the
218 IMC-PGRF and OMC-PGRF were compared to OMC-MGRF.

219 Forces were normalized to body weight (BW) and moments to body weight
220 times body height (BW*BH). The time axis of the curves was normalized to
221 100% of the gait cycle for the kinematics (time between two consecutive heel-
222 strike events of the analyzed limb) and 100% of the stance phase (time between
223 heel-strike and toe-off events of the analyzed limb) for the kinetics. Measured
224 and estimated GRF&M were expressed on the right handed coordinate frame
225 defined by the walking direction within the trial (given that the subjects walked
226 straight) and the vertical axis equal to the vertical axis of the respective mo-
227 tion capture system used. On the other hand, JRF&M were expressed on the
228 coordinate frame of the segment distal to the body in both IMC and OMC
229 methods.

230 The above-mentioned comparisons of kinematic and kinetic variables to their
231 respective references were performed using absolute and relative root-mean-
232 square-differences (RMSD and rRMSD, respectively) as described by Ren *et al.*
233 [19]. In addition, for every curve, the magnitude (M) and phase (P) differ-
234 ence metrics [41] have been utilized. Pearson correlation coefficient (ρ) were
235 calculated, averaged using Fisher's z transformation method [42], and cate-
236 gorized similarly to Taylor *et al.* [43], as "weak" ($\rho \leq 0.35$), "moderate"
237 ($0.35 < \rho \leq 0.67$), "strong" ($0.67 < \rho \leq 0.90$), and "excellent" ($\rho > 0.90$).

238 3. Results

239 3.1. Estimated kinematics of the musculoskeletal model

240 Table 1 presents the results for the accuracy analysis for the joint angles
241 of the IMC-driven model versus the OMC-driven model. Similarly, Figure 2
242 illustrates the curves for the joint angles of the lower extremities averaged across
243 all gait cycles performed by the eleven subjects. Excellent Pearson correlation
244 coefficients have been found in all sagittal plane angles for ankle, knee, and hip
245 (0.95, 0.99, and 0.99, respectively). For the same variables, the RMSDs across a
246 gait cycle were found as $4.1 \pm 1.3^\circ$, $4.4 \pm 2.0^\circ$ and $5.7 \pm 2.1^\circ$, respectively (mean
247 \pm standard deviation). Hip flexion angles were overall underestimated ($M =$

248 $-4.0 \pm 13.9\%$), whereas knee and ankle magnitude differences showed an average
249 overestimation ($0.7 \pm 6.2\%$ and $8.6 \pm 16.4\%$). The hip abduction showed excellent
250 correlations ($\rho = 0.91$) with an RMSD of $4.1 \pm 2.0^\circ$ and a mean underestimation
251 with a magnitude difference $M = -12.2 \pm 34.7\%$. Strong correlation values ($\rho =$
252 0.68) were observed in the hip internal-external rotation angle with an RMSD
253 of $6.5 \pm 2.8^\circ$ and an underestimation of magnitude difference $M = 5.5 \pm 39.0\%$.
254 Finally, the subtalar eversion angle showed strong correlation ($\rho = 0.82$), RMSD
255 of $9.66 \pm 3.07^\circ$ and $M = 24.0 \pm 34.7\%$.

256 *3.2. Predicted kinetics using inertial and optical motion capture*

257 The results of the accuracy analysis for GRF&M and JRF&M are presented
258 in Table 2 and 3, for IMC-PGRF and OMC-PGRF, respectively. The mean
259 values and standard deviations of the curves from IMC-PGRF, OMC-PGRF,
260 and OMC-MGRF models, are illustrated in Figures 3 and 4, for the forces and
261 moments, respectively.

262 The Pearson correlation coefficients of the IMC-PGRF model were excellent
263 for vertical ($\rho = 0.97$) and anteroposterior GRF&M ($\rho = 0.91$) and strong for
264 mediolateral GRF&M ($\rho = 0.80$). For the same components, RMSD values
265 observed were of 9.3 ± 3.0 , 5.5 ± 1.2 and 2.1 ± 0.6 %BW, respectively (mean
266 \pm standard deviation). The OMC-PGRF model performed better in the an-
267 teroposterior GRF&M components ($\rho = 0.96$, RMSD = 3.7 ± 1.1 %BW), and
268 similarly to IMC-PGRF for the other two GRF&M components (mediolateral:
269 $\rho = 0.79$, RMSD = 1.9 ± 0.5 BW, vertical: $\rho = 0.99$, RMSD = 5.9 ± 1.9 BW).

270 Concerning GRM, the sagittal plane was predicted with similar excellent
271 correlations in both IMC-PGRF ($\rho = 0.91$) and OMC-PGRF ($\rho = 0.94$) driven
272 models. The correlation coefficients for frontal and transverse GRM components
273 found in the IMC-PGRF model were $\rho = 0.64$, $\rho = 0.82$, respectively, whereas
274 in the OMC-PGRF model ($\rho = 0.66$, $\rho = 0.81$, respectively). The RMSDs
275 found in the IMC-PGRF approach were 0.9 ± 0.6 , 1.6 ± 0.6 , and 0.2 ± 0.001
276 %BW*BH for frontal, sagittal and transverse GR&M, respectively, which were
277 either slightly higher or similar to the RMSDs of the OMC-PGRF approach

278 (0.7 ± 0.2, 1.2 ± 0.4, and 0.2 ± 0.1 %BW*BH, respectively).

279 4. Discussion

280 We have presented a method to perform musculoskeletal model-based in-
281 verse dynamics using exclusively IMC input (IMC-PGRF). First, we compared
282 the kinematic joint angle estimates of the lower limbs against those assessed
283 through a conventional, laboratory-based OMC input. In addition, we tested
284 the performance of the approach in calculating the JRF&M, while predicting
285 the GRF&M from the kinematics, against a similarly constructed model (OMC-
286 MGRF) which uses input from both FP and OMC. Finally, we performed a sim-
287 ilar comparison to evaluate the predicted kinetics of a model driven exclusively
288 by OMC input (OMC-PGRF).

289 Regarding the IMC-based joint angles in the musculoskeletal model, all three
290 sagittal plane angles provided excellent correlations (range: 0.95-0.99) and aver-
291 age RMSD values remained below 6°. Slightly lower correlations were observed
292 in the frontal and transverse plane angles, which can be explained due to the
293 smaller range of motion within these planes. For instance, even though the
294 hip abduction and external rotation joint angles present absolute RMSD values
295 similar to the flexion component, their rRMSDs which take into account the
296 range of motion are two and three times higher, respectively.

297 Both GRF&M and JRF&M of the vertical axis presented higher correlations
298 and lower RMSDs than the ones in the anteroposterior and mediolateral axes.
299 Similarly, sagittal plane moments were found in most cases to be more accurate
300 than frontal and transverse plane moments. By visual inspection of the curves,
301 we observe that the magnitude of the IMC-PGRF anteroposterior GRF&M
302 seems to be underestimated both in the negative early stance and positive late
303 stance peak, which can be confirmed by the magnitude difference for that curve
304 ($M = -28.3\%$). However, this behaviour is not observed in the OMC-PGRF,
305 nor during the single stance of the IMC-PGRF curve. Despite the higher rRMSD
306 found in the non-sagittal joint angles, the performance of the IMC-PGRF in

307 the mediolateral, frontal and transverse plane GRF&M components matched
308 closely the OMC-PGRF approach. This observation reveals that OMC-based
309 kinematics suffer from errors of similar size, when capturing the typically small
310 movements of the frontal and transverse planes, given the fact that both IMC-
311 PGRF and OMC-PGRF had the same model characteristics. Therefore, OMC-
312 MGRF should also be used with caution, when comparing either kinematic or
313 JRF&M quantities of the non-sagittal planes.

314 A number of error sources contribute to discrepancies in the OMC kinemat-
315 ics. First, soft tissue artefacts can create a relative movement of the marker
316 with respect to the bone [44, 45]. In addition, mismatches between the experi-
317 mental and modelled marker positions can lead to errors in segment orientations
318 calculated during inverse kinematics. Both error sources would have a relatively
319 larger impact on the kinematics of the frontal and transverse plane, than on the
320 sagittal plane. Finally, the JRF&M of the OMC-PGRF were compared against
321 a non-independent OMC-MGRF reference, which could have contributed to un-
322 derestimation of the actual errors.

323 The IMC-PGRF approach has a number of possible sources of errors which
324 would influence the performance. Similarly to OMC models, soft-tissue ar-
325 tifacts may compromise the kinematic estimates. Further errors in segment
326 kinematics may stem due to the N-pose calibration assumptions. In particular,
327 mismatches between the practised and modelled N-pose could result in offsets
328 in the estimated positions. Other common error sources in IMC include manual
329 measurements of segment lengths as well as IMU inaccuracies. In addition, the
330 stick figure model, which was utilized to recreate the VMs, has a higher number
331 of DOF, compared to the musculoskeletal model used.

332 A possible source of error in all inverse dynamic approaches concerns the
333 inertial parameters (masses and moments of inertia), as well as the center of
334 mass (CoM) locations of each human body segment, which were calculated
335 based on anthropometric tables found in the literature.

336 This study focused on presenting and evaluating a general workflow for mus-
337 culoskeletal model-based inverse dynamic simulations using ambulatory IMC

338 systems. The presentation of results in this study was performed on the level of
339 ground and joint reaction forces and moments. These measures are calculated
340 from muscle force estimates derived from a muscle recruitment optimization
341 technique. Given the high number of muscles in the model (110) and without
342 a clear medical research question, it is challenging to choose which muscles are
343 more important to present and analyze. Future studies could examine specific
344 applications and pathologies in order to identify the most important muscles
345 and evaluate their respective force estimates.

346 A limitation of this study is that, even though the method has been pre-
347 viously shown to be universally applicable in OMC-based studies [22, 23], we
348 only evaluated its performance in gait of three different speeds. In addition,
349 our experiments included only young healthy male subjects, but the underlying
350 methods to predict kinetics from kinematics have been recently shown to be
351 applicable in Parkinson’s patients [46]. Future studies could investigate the ap-
352 plication of IMC systems combined with musculoskeletal modeling in groups of
353 larger sample size than the current study, including patients, as well as female
354 subjects.

355 **5. Conclusion**

356 In this study, we have demonstrated a workflow to perform musculoskeletal
357 model-based inverse dynamics using input from a commercially available IMC
358 system. Our validation findings indicate that the prediction of GRF&M as well
359 as JRF&M using musculoskeletal model-based inverse dynamics based on only
360 IMC data provides comparable performance to both OMC-PGRF and OMC-
361 MGRF methods. The proposed method allows assessment of kinetic variables
362 outside the laboratory.

363 **Ethical approval**

364 The ethical guidelines of The Scientific Ethical Committee for the Region of
365 North Jutland (Den Videnskabetiske Komit for Region Nordjylland) were fol-

366 lowed and all volunteers signed written informed consent after receiving detailed
367 information prior to data collection.

368 **Conflict of interest statement**

369 Three of the authors are employees of Xsens Technologies B.V. that manu-
370 factures and sells the Xsens MVN. One of the authors is employee of AnyBody
371 Technology A/S that owns and sells the AnyBody Modeling System.

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546 **List of Figures**

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548 recording from Xsens MVN Studio (a) is exported to a BVH
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551 of the musculoskeletal model (c), and by projecting b on c the
552 kinematics of the musculoskeletal model are solved. Finally, in-
553 verse dynamic analysis using prediction of ground reaction forces
554 and moments is performed to estimate the kinetics. 24

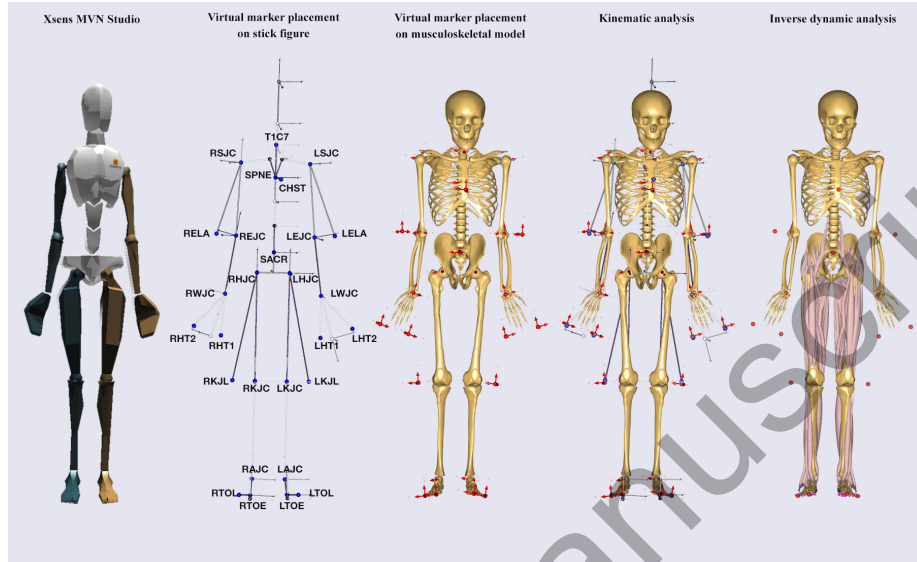
555 2 Ankle, knee, and hip joint angle estimates (standard deviation
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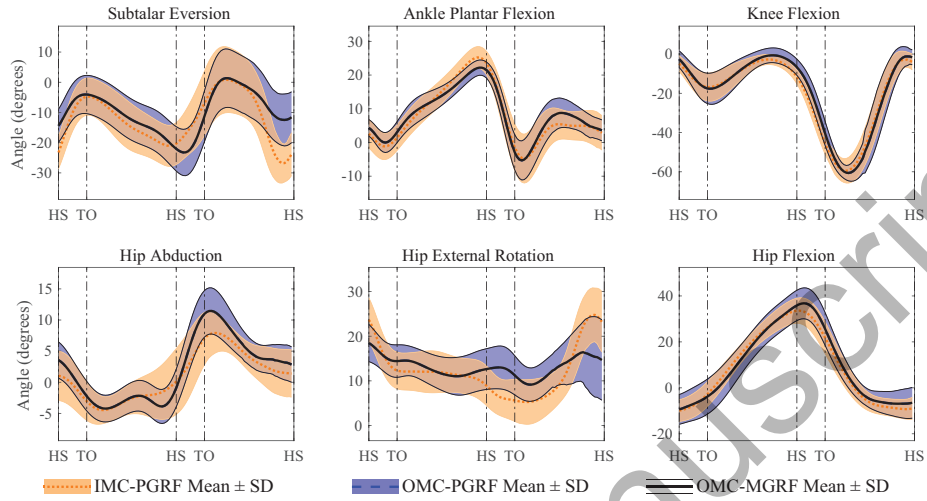
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Figure 1



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Figure 2



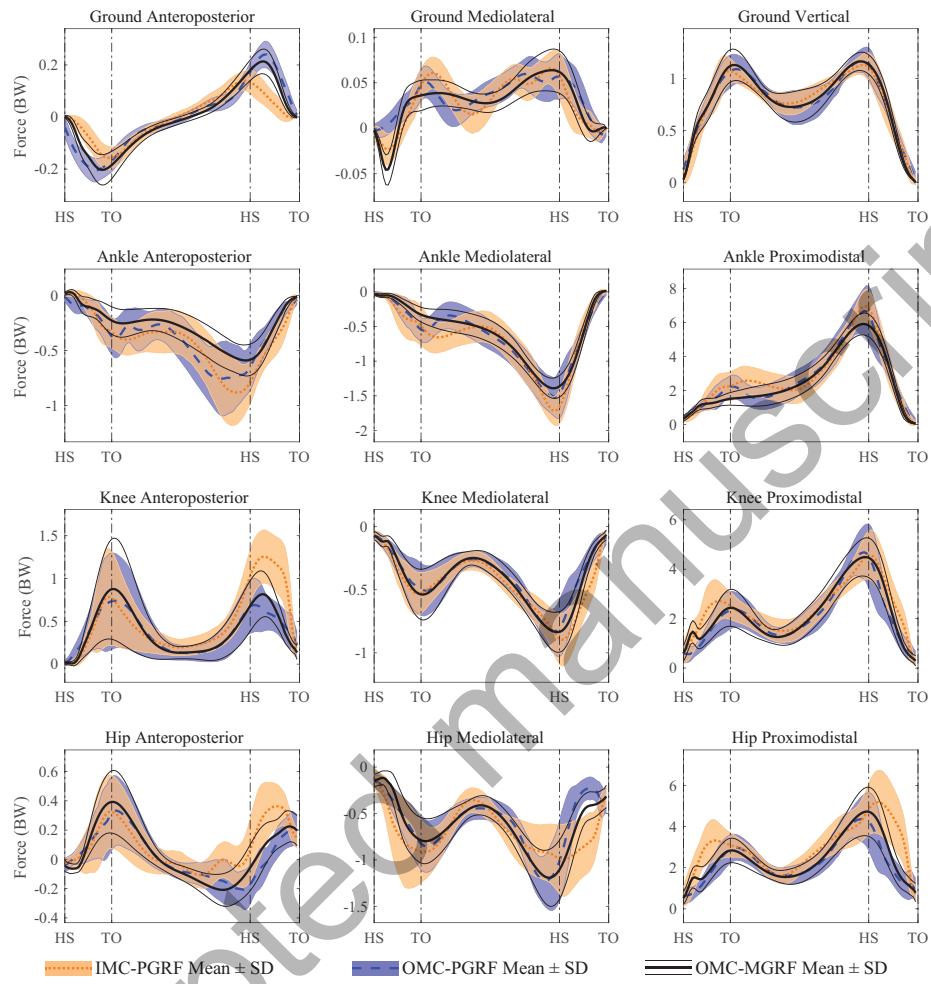


Figure 3

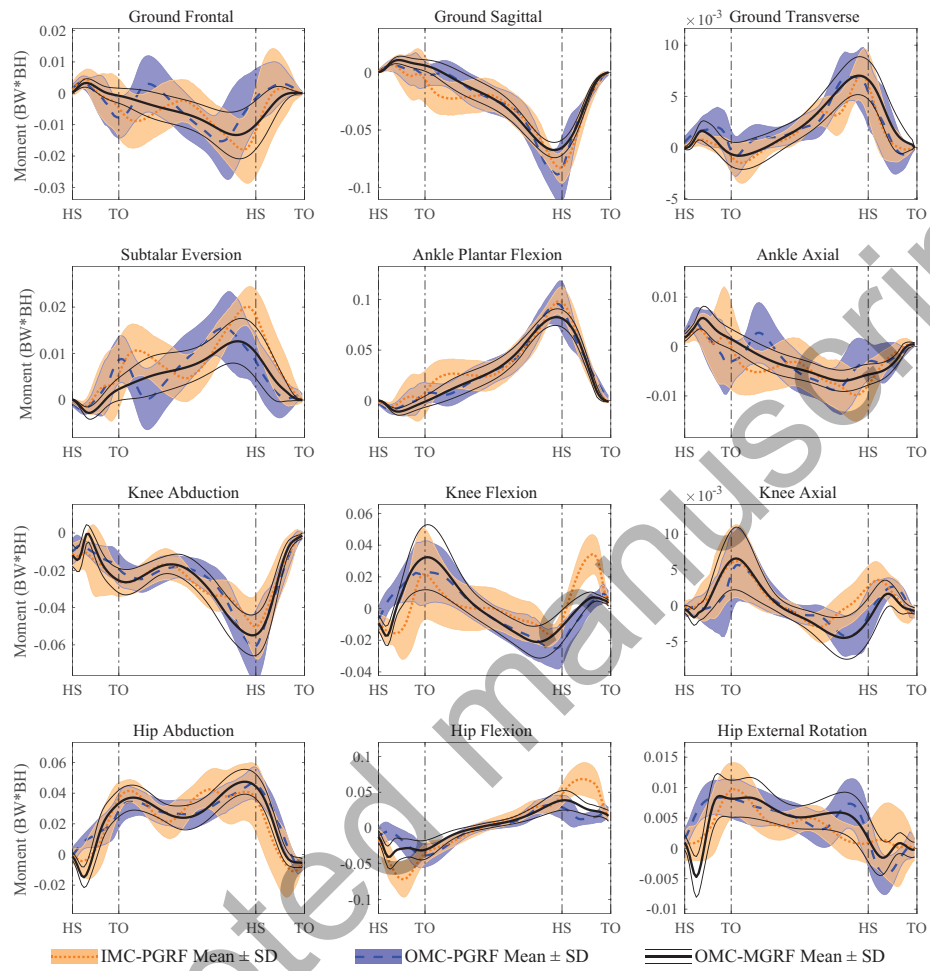


Figure 4

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576 M and P denote the % magnitude and phase differences 29

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579 OMC-MGRF. Pearson correlation coefficient is denoted with ρ .
580 Absolute per body weight (or body weight times height) and
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582 (%BW or %BW*BH) and $rRMSD$ (%), respectively. M and P
583 indicate the magnitude and phase differences (%). 30

584 3 OMC-PGRF-based ground and joint reaction forces (first three
585 quantities) and net moments (second three quantities) versus
586 OMC-MGRF. Pearson correlation coefficient is denoted with ρ .
587 Absolute per body weight (or body weight times height) and
588 relative root-mean-squared-difference are denoted with $RMSD$
589 (%BW or %BW*BH) and $rRMSD$ (%), respectively. M and P
590 indicate the magnitude and phase differences (%). 31

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Table 1

	ρ	RMSD	rRMSD	M	P
Subtalar Eversion	0.81	9.7 (3.2)	32.6 (10.3)	24.0 (34.7)	19.3 (10.2)
Ankle Plantarflexion	0.95	4.1 (1.3)	14.0 (4.8)	8.6 (16.4)	9.8 (3.9)
Knee Flexion	0.99	4.4 (2.0)	7.2 (3.4)	0.7 (6.2)	4.8 (2.7)
Hip Abduction	0.91	4.1 (2.0)	25.9 (10.7)	-12.2 (34.7)	21.2 (9.3)
Hip External Rotation	0.68	6.5 (2.8)	36.9 (15.2)	5.5 (39.0)	12.6 (6.2)
Hip Flexion	0.99	5.7 (2.1)	12.7 (5.3)	-4.0 (13.9)	8.8 (4.2)

Table 2

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.91	5.5 (1.2)	15.0 (2.4)	-25.4 (7.3)	14.4 (3.2)
Mediolateral	0.80	2.1 (0.6)	18.5 (3.2)	7.3 (19.3)	15.4 (3.8)
Vertical	0.97	9.3 (3.0)	7.7 (2.1)	-1.5 (1.5)	3.4 (1.0)
Frontal	0.64	0.9 (0.6)	38.0 (23.1)	125.5 (319.9)	30.6 (17.3)
Sagittal	0.91	1.6 (0.6)	17.5 (6.8)	14.3 (18.2)	12.1 (4.5)
Transverse	0.82	0.2 (0.1)	23.3 (7.2)	-8.5 (41.9)	17.8 (5.3)
Ankle					
Anteroposterior	0.84	22.2 (10.3)	26.1 (10.2)	49.0 (45.8)	10.8 (2.1)
Mediolateral	0.93	24.3 (8.9)	15.2 (5.3)	14.3 (17.1)	7.9 (2.7)
Proximodistal	0.93	88.5 (30.6)	13.6 (4.6)	9.8 (14.1)	7.2 (2.3)
Eversion	0.76	0.6 (0.2)	33.3 (20.2)	107.7 (220.3)	18.9 (10.7)
Plantar Flexion	0.93	1.6 (0.6)	15.1 (6.6)	10.6 (18.1)	9.9 (3.6)
Axial	0.67	0.5 (0.2)	30.4 (12.2)	46.5 (49.1)	27.2 (13.5)
Knee					
Anteroposterior	0.82	30.6 (10.3)	25.8 (9.7)	43.7 (53.5)	13.0 (4.5)
Mediolateral	0.91	12.0 (3.5)	14.1 (3.8)	6.6 (8.6)	7.0 (2.0)
Proximodistal	0.90	63.1 (26.9)	14.3 (6.6)	5.1 (9.1)	7.2 (2.8)
Abduction	0.81	1.1 (0.4)	18.9 (6.8)	-2.7 (16.1)	10.7 (3.8)
Flexion	0.58	1.9 (0.5)	29.8 (7.6)	17.9 (45.0)	32.8 (9.6)
Axial	0.73	0.3 (0.1)	25.4 (10.3)	2.3 (30.5)	27.9 (13.8)
Hip					
Anteroposterior	0.71	17.6 (7.6)	27.2 (9.6)	6.8 (24.4)	27.6 (10.9)
Mediolateral	0.73	27.0 (12.5)	23.0 (7.4)	7.7 (14.6)	10.6 (4.1)
Proximodistal	0.78	102.8 (30.6)	21.7 (4.5)	20.2 (10.0)	9.0 (2.5)
Abduction	0.83	1.4 (0.7)	19.7 (5.8)	6.3 (16.9)	13.7 (7.9)
Flexion	0.92	2.2 (0.6)	19.4 (4.2)	73.2 (26.3)	14.8 (4.2)
External Rotation	0.50	0.5 (0.2)	31.6 (6.6)	-3.9 (36.4)	25.6 (10.1)

Table 3

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.96	3.7 (1.1)	8.3 (2.0)	7.7 (12.0)	8.8 (1.8)
Mediolateral	0.79	1.9 (0.5)	18.6 (4.1)	2.4 (10.8)	15.2 (4.9)
Vertical	0.99	5.9 (1.9)	4.9 (1.4)	-1.2 (1.1)	2.1 (0.7)
Frontal	0.66	0.7 (0.2)	30.3 (9.3)	71.0 (122.2)	24.5 (9.1)
Sagittal	0.94	1.2 (0.4)	13.1 (3.8)	15.9 (15.3)	9.2 (3.2)
Transverse	0.81	0.2 (0.1)	20.7 (7.5)	7.1 (22.9)	17.5 (7.5)
Ankle					
Anteroposterior	0.83	18.9 (6.9)	23.0 (6.1)	37.3 (28.6)	10.8 (2.3)
Mediolateral	0.96	16.1 (4.2)	10.7 (2.6)	6.8 (9.6)	5.8 (2.1)
Proximodistal	0.96	62.2 (17.6)	9.8 (2.7)	7.1 (9.0)	5.2 (1.8)
Eversion	0.76	0.5 (0.1)	25.5 (7.0)	45.3 (64.1)	18.7 (10.2)
Plantar Flexion	0.96	1.0 (0.3)	10.1 (3.3)	5.9 (10.0)	7.0 (2.6)
Axial	0.64	0.5 (0.1)	27.2 (7.3)	33.3 (36.9)	27.5 (11.5)
Knee					
Anteroposterior	0.93	11.9 (4.5)	12.3 (4.3)	-7.3 (8.7)	7.4 (2.0)
Mediolateral	0.96	7.2 (2.0)	8.8 (2.6)	-4.2 (5.6)	4.4 (1.0)
Proximodistal	0.95	41.7 (12.0)	9.3 (2.6)	-2.7 (5.8)	4.9 (1.2)
Abduction	0.91	0.8 (0.2)	12.6 (2.6)	-0.1 (10.5)	7.7 (1.6)
Flexion	0.86	0.9 (0.3)	16.7 (4.8)	-1.7 (14.3)	16.9 (5.2)
Axial	0.82	0.2 (0.1)	18.5 (6.6)	-3.4 (17.7)	20.6 (8.0)
Hip					
Anteroposterior	0.89	9.9 (3.6)	16.0 (6.7)	-10.4 (10.6)	16.6 (7.6)
Mediolateral	0.92	14.7 (4.0)	12.7 (3.1)	-1.9 (6.9)	6.2 (1.5)
Proximodistal	0.92	50.0 (15.9)	11.5 (2.6)	-4.6 (6.1)	5.5 (1.2)
Abduction	0.91	0.8 (0.2)	13.3 (2.6)	-3.2 (6.3)	8.7 (2.4)
Flexion	0.86	1.3 (0.4)	16.4 (3.4)	-9.3 (12.3)	18.0 (4.1)
External Rotation	0.68	0.3 (0.1)	22.5 (3.7)	6.5 (15.8)	18.8 (4.8)

Supplementary material

Title:

Musculoskeletal model-based inverse dynamic analysis under ambulatory conditions using inertial motion capture

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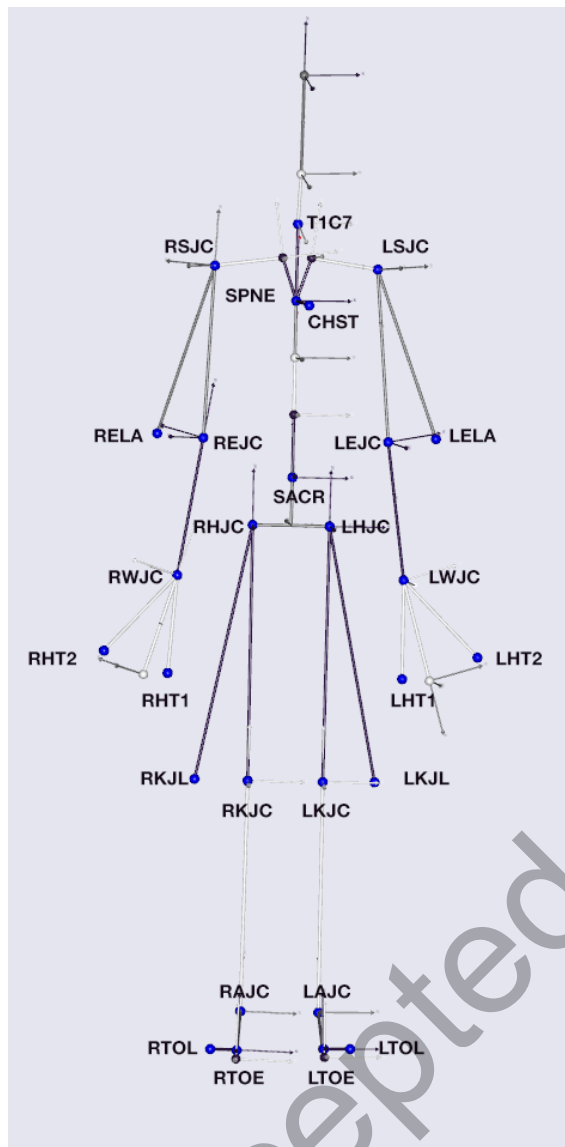
Keywords:

musculoskeletal modeling, inertial motion capture, inverse dynamics, ground reaction forces and moments, gait analysis

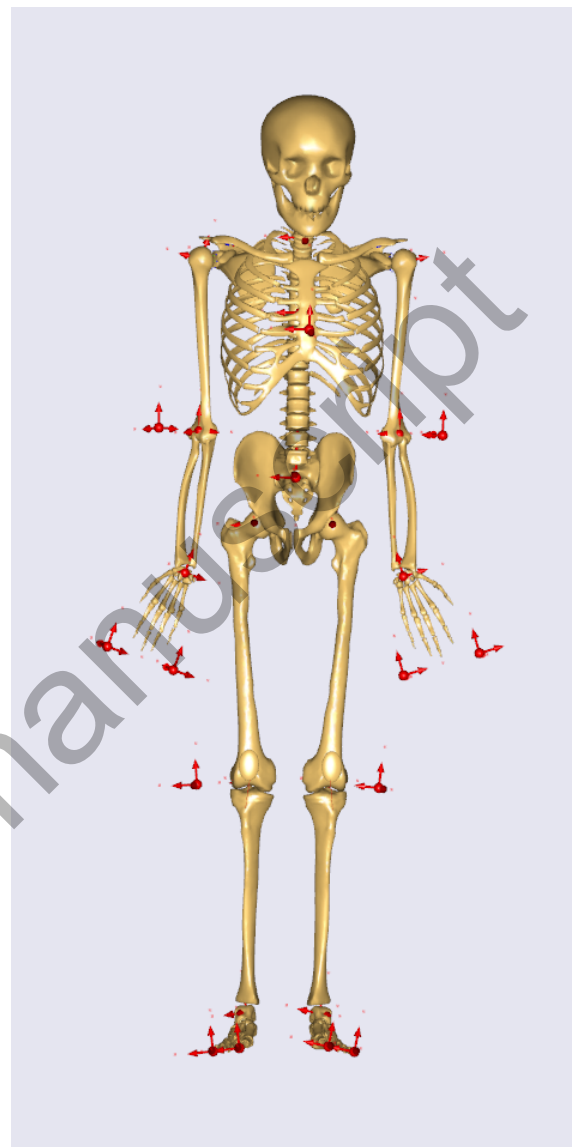
1. Virtual marker placement

Table 1: Description of the placement of virtual markers (VM) on the segments of the Xsens MVN model (stick figure model) and the musculoskeletal model constructed based on the AnyBody Managed Model Repository (AMMR).

VM Name	VM Placement on MVN	VM Placement on AMMR	VM Weight
T1C7	jT1C7	T1/C7 Joint	(1,1,1)
SPNE	jT9T8	Inferior to T1/C7 Joint	(1,1,1)
CHST	Anterior to jT9T8	Inferior and Anterior to T1/C7 Joint	(1,1,1)
SACR	jL5S1	Anterior to Pelvis/Sacrum Joint	(10,0,0)
RHJC	jRightHip	Right Hip Joint	(10,10,10)
RKJC	jRightKnee	Right Knee Joint	(2,2,2)
RKJL	Lateral to jRightKnee	Lateral to Right Knee Joint	(1,0,0)
RAJC	jRightAnkle	Right Ankle Joint	(1,1,1)
RTOE	jRightBallFoot	Right Big Toe Node	(1,1,1)
RTOL	Lateral to jRightBallFoot	Lateral to Right Big Toe Node	(0,1,0)
RSJC	jRightShoulder	Right Glenohumeral Joint	(0,2,2)
REJC	jRightElbow	Elbow Joint	(2,2,2)
RELA	Lateral to jRightElbow	Lateral to Elbow Joint	(1,1,1)
RWJC	jRightWrist	Right Wrist Joint	(2,2,2)
RHT1	Inferior and Medial to jRightWrist	Inferior and Medial to Right Wrist Joint	(0.5,0.5,0.5)
RHT2	Inferior and Lateral to jRightWrist	Inferior and Lateral to Right Wrist Joint	(0.5,0.5,0.5)
LHJC	jLeftHip	Left Hip Joint	(10,10,10)
LKJC	jLeftKnee	Left Knee Joint	(2,2,2)
LKJL	Lateral to jLeftKnee	Lateral to Left Knee Joint	(1,0,0)
LAJC	jLeftAnkle	Left Ankle Joint	(1,1,1)
LTOE	jLeftBallFoot	Left Big Toe Node	(1,1,1)
LTOL	Lateral to jLeftBallFoot	Lateral to Left Big Toe Node	(0,1,0)
LSJC	jLeftShoulder	Left Glenohumeral Joint	(0,2,2)
LEJC	jLeftElbow	Elbow Joint	(2,2,2)
LELA	Lateral to jLeftElbow	Lateral to Elbow Joint	(1,1,1)
LWJC	jLeftWrist	Left Wrist Joint	(2,2,2)
LHT1	Inferior and Medial to jLeftWrist	Inferior and Medial to Left Wrist Joint	(0.5,0.5,0.5)
LHT2	Inferior and Lateral to jLeftWrist	Inferior and Lateral to Left Wrist Joint	(0.5,0.5,0.5)



(a) Xsens MVN (stick-figure) model



(b) AnyBody Musculoskeletal Model (AMMR)

Figure 1: Illustration of the placement of the virtual markers (VM) on the segments of the Xsens MVN model (stick figure model) and the musculoskeletal model constructed based on the AnyBody Managed Model Repository (AMMR).

2. Accuracy analysis per walking speed

2.1. Comfortable walking speed

Table 2: Comfortable walking speed; comparison of lower limb joint angles between musculoskeletal model driven by the inertial (IMC-PGRF) and optical motion capture (OMC-PGRF/OMC-MGRF), using Pearson correlation coefficient (ρ), absolute and relative root-mean-squared-differences ($RMSD$ in degrees and $rRMSD$ in %, respectively). M and P denote the % magnitude and phase differences.

Normal Walking	ρ	$RMSD$	$rRMSD$	M	P
Subtalar Eversion	0.79	9.7 (3.1)	32.6 (10.1)	25.5 (36.2)	18.9 (9.6)
Ankle Plantarflexion	0.95	4.0 (1.3)	13.1 (4.9)	10.3 (16.6)	9.3 (3.6)
Knee Flexion	0.98	4.6 (2.0)	7.4 (3.1)	2.1 (5.5)	4.9 (2.3)
Hip Abduction	0.91	3.9 (1.9)	25.2 (9.1)	-15.9 (28.8)	20.4 (8.0)
Hip External Rotation	0.66	6.5 (2.6)	35.7 (14.1)	7.9 (36.7)	12.6 (5.5)
Hip Flexion	0.99	5.6 (2.2)	12.5 (5.5)	-3.7 (13.0)	8.8 (4.4)

Figure 2: Comfortable walking speed; ankle, knee, and hip joint angle estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

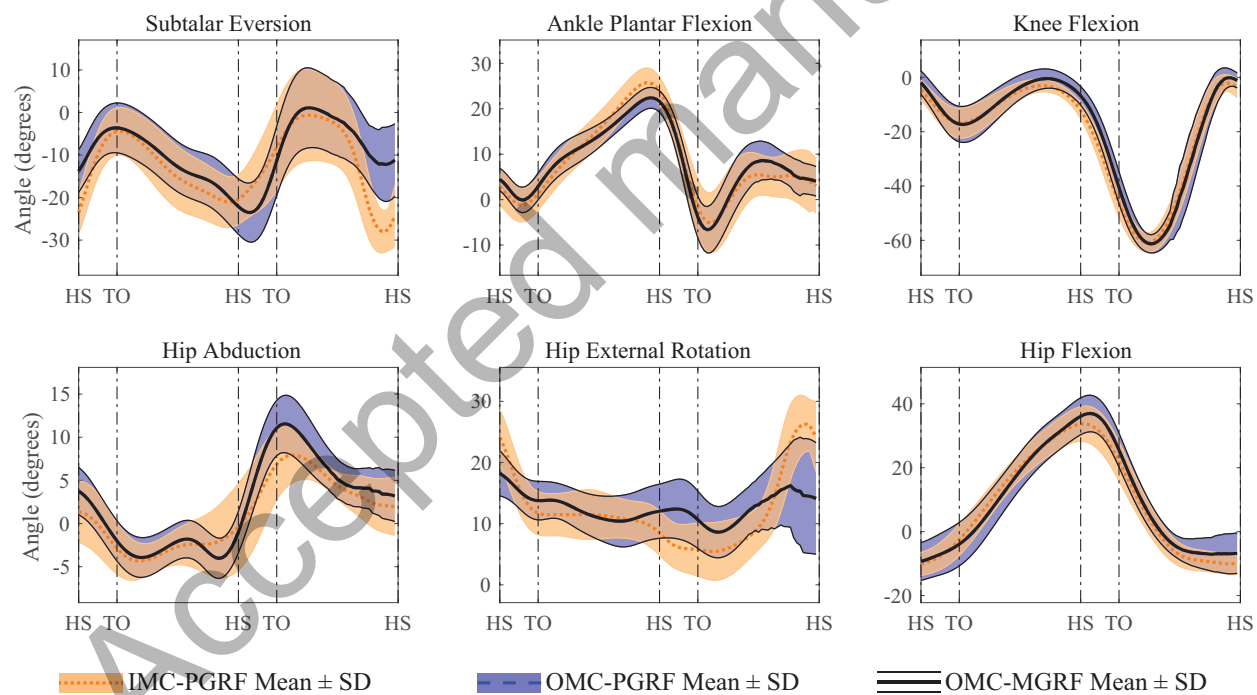


Table 3: Comfortable walking speed; IMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with *RMSD* (%BW or %BW*BH) and *rRMSD* (%), respectively. *M* and *P* indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.91	5.5 (1.1)	14.6 (2.4)	-24.8 (6.6)	14.2 (3.0)
Mediolateral	0.80	2.1 (0.5)	18.7 (2.5)	4.8 (17.4)	15.5 (3.0)
Vertical	0.97	8.6 (2.3)	7.1 (1.8)	-1.4 (1.4)	3.1 (0.8)
Frontal	0.66	0.8 (0.5)	34.1 (15.3)	105.5 (334.8)	28.5 (13.8)
Sagittal	0.91	1.5 (0.6)	16.8 (7.0)	12.2 (17.2)	11.6 (4.4)
Transverse	0.83	0.2 (0.1)	22.3 (6.4)	-11.7 (31.7)	17.4 (4.3)
Ankle					
Anteroposterior	0.84	21.9 (10.3)	25.7 (10.1)	48.0 (47.2)	10.9 (2.1)
Mediolateral	0.93	23.9 (8.3)	14.8 (4.9)	12.8 (15.4)	8.0 (2.5)
Proximodistal	0.93	85.6 (27.7)	13.0 (4.4)	8.0 (12.6)	7.2 (2.2)
Eversion	0.75	0.6 (0.2)	31.5 (16.3)	98.7 (234.1)	18.3 (8.6)
Plantar Flexion	0.94	1.5 (0.6)	14.3 (6.0)	8.2 (15.3)	9.6 (3.5)
Axial	0.70	0.5 (0.2)	29.2 (11.6)	38.7 (46.6)	25.4 (12.8)
Knee					
Anteroposterior	0.84	29.8 (9.3)	26.0 (9.8)	49.4 (50.6)	12.3 (4.7)
Mediolateral	0.93	11.4 (3.0)	13.5 (3.8)	8.1 (8.7)	6.4 (2.2)
Proximodistal	0.92	58.4 (29.2)	13.3 (6.9)	5.1 (8.1)	6.7 (3.1)
Abduction	0.83	1.0 (0.4)	17.6 (6.4)	-2.1 (14.7)	10.3 (4.0)
Flexion	0.59	1.8 (0.5)	29.8 (6.8)	16.7 (38.8)	32.9 (8.2)
Axial	0.73	0.3 (0.2)	25.4 (9.9)	2.7 (31.6)	27.8 (13.1)
Hip					
Anteroposterior	0.74	16.7 (7.4)	26.2 (8.8)	5.7 (17.7)	26.6 (8.8)
Mediolateral	0.75	26.1 (13.4)	22.7 (7.6)	7.1 (15.4)	10.3 (4.2)
Proximodistal	0.81	99.0 (25.3)	21.6 (4.4)	21.2 (10.1)	8.5 (2.7)
Abduction	0.84	1.3 (0.7)	18.8 (5.5)	9.5 (17.9)	12.8 (8.2)
Flexion	0.92	2.2 (0.6)	18.9 (3.5)	69.5 (20.9)	14.6 (4.6)
External Rotation	0.47	0.5 (0.2)	30.9 (6.9)	-6.7 (32.6)	25.6 (9.9)

Table 4: Comfortable walking speed; OMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with $RMSD$ (%BW or %BW*BH) and $rRMSD$ (%), respectively. M and P indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.96	3.6 (1.0)	8.2 (1.8)	5.8 (8.5)	8.9 (1.8)
Mediolateral	0.77	1.9 (0.4)	19.1 (3.7)	1.6 (10.8)	15.4 (4.1)
Vertical	0.99	5.7 (1.2)	4.8 (1.0)	-1.2 (0.9)	2.1 (0.5)
Frontal	0.66	0.7 (0.2)	30.9 (9.4)	68.9 (139.9)	24.7 (8.2)
Sagittal	0.94	1.1 (0.2)	12.0 (2.6)	15.8 (11.3)	8.4 (2.0)
Transverse	0.82	0.2 (0.1)	19.8 (7.2)	3.8 (21.4)	16.6 (6.4)
Ankle					
Anteroposterior	0.85	18.8 (6.5)	23.2 (6.6)	40.1 (29.7)	10.3 (2.1)
Mediolateral	0.96	15.0 (2.9)	9.8 (1.9)	6.7 (7.5)	5.5 (1.6)
Proximodistal	0.97	57.2 (12.6)	8.8 (2.0)	7.0 (6.7)	4.8 (1.3)
Eversion	0.75	0.5 (0.1)	25.5 (6.8)	45.4 (75.0)	18.0 (8.7)
Plantar Flexion	0.97	0.9 (0.2)	8.9 (2.0)	5.7 (7.3)	6.3 (1.7)
Axial	0.63	0.5 (0.1)	27.3 (6.5)	29.9 (35.3)	26.6 (10.1)
Knee					
Anteroposterior	0.93	11.0 (4.3)	11.2 (2.7)	-6.9 (5.4)	7.2 (1.4)
Mediolateral	0.97	6.5 (1.6)	8.0 (1.8)	-4.2 (3.8)	4.1 (0.7)
Proximodistal	0.96	37.7 (7.8)	8.4 (1.7)	-2.7 (4.0)	4.6 (0.9)
Abduction	0.91	0.7 (0.1)	11.9 (2.3)	-0.3 (8.1)	7.8 (1.7)
Flexion	0.86	0.9 (0.2)	16.7 (4.5)	-1.2 (12.9)	17.2 (5.2)
Axial	0.81	0.2 (0.1)	18.7 (6.2)	-6.5 (17.2)	20.6 (6.9)
Hip					
Anteroposterior	0.89	9.6 (3.0)	15.6 (6.2)	-11.8 (9.3)	16.8 (7.0)
Mediolateral	0.91	14.7 (3.2)	12.6 (2.2)	-2.2 (7.0)	6.4 (1.0)
Proximodistal	0.92	47.5 (13.1)	11.2 (2.2)	-4.9 (4.9)	5.4 (1.0)
Abduction	0.90	0.8 (0.1)	13.3 (2.4)	-3.6 (5.7)	8.9 (2.0)
Flexion	0.86	1.3 (0.3)	16.0 (2.6)	-9.0 (11.8)	17.6 (3.4)
External Rotation	0.67	0.3 (0.1)	22.7 (3.5)	7.0 (15.7)	18.8 (4.4)

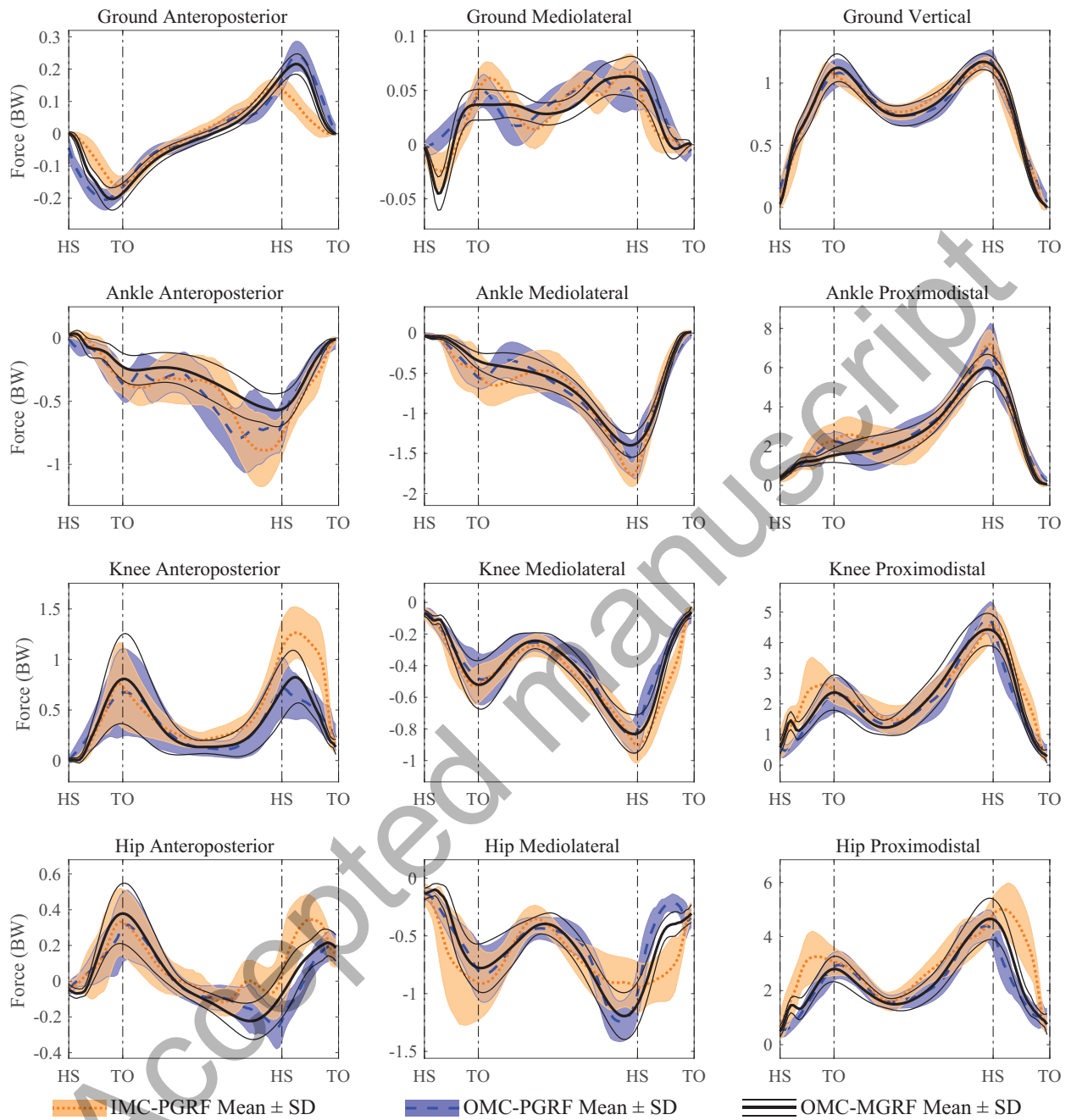


Figure 3: Comfortable walking speed; ground and lower limb joint reaction force estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

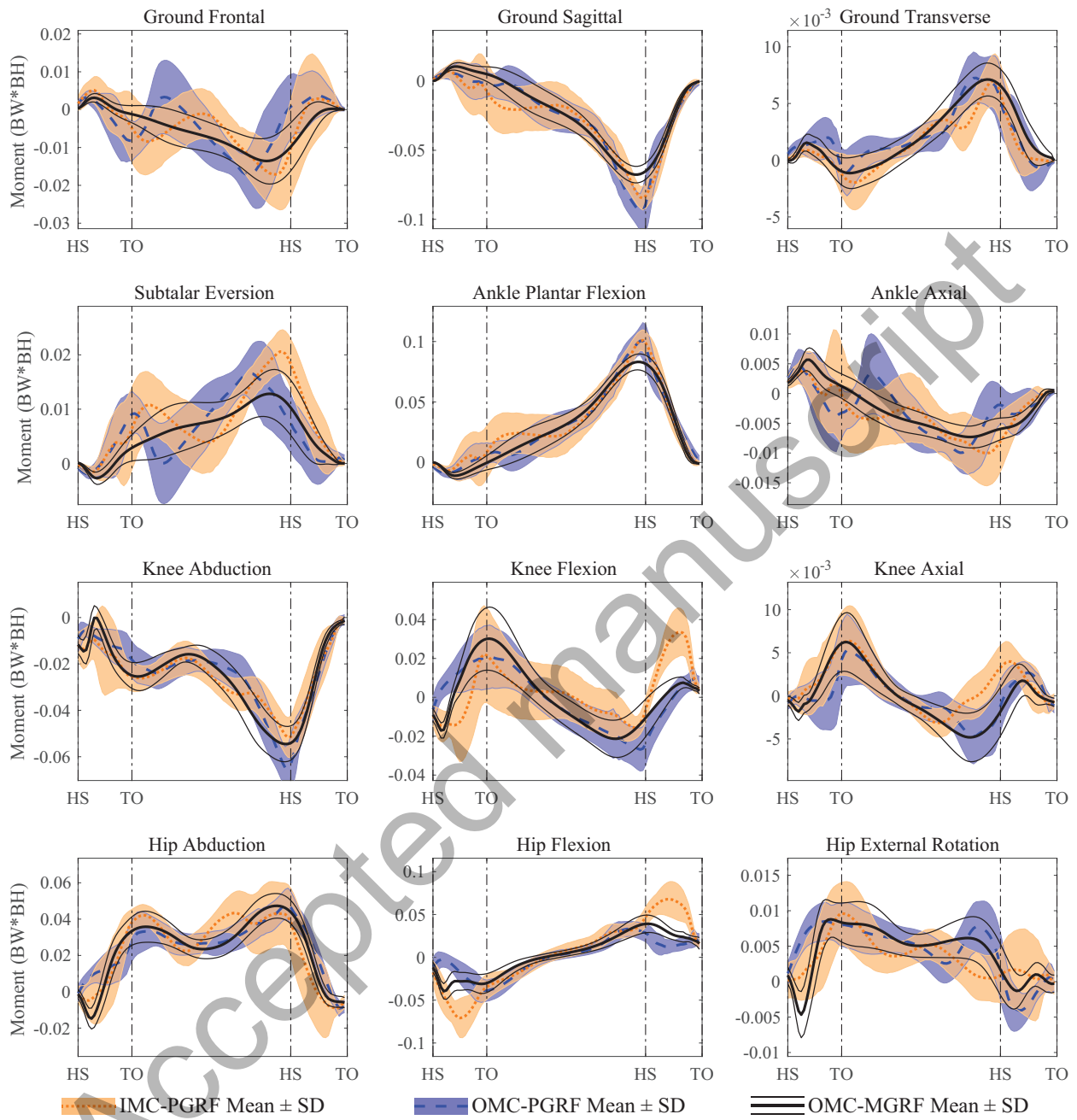


Figure 4: Comfortable walking speed; ground reaction and lower limb net joint moment estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

2.2. Slow walking speed

Table 5: Slow walking speed; comparison of lower limb joint angles between musculoskeletal model driven by the inertial (IMC-PGRF) and optical motion capture (OMC-PGRF/OMC-MGRF), using Pearson correlation coefficient (ρ), absolute and relative root-mean-squared-differences ($RMSD$ in degrees and $rRMSD$ in %, respectively). M and P denote the % magnitude and phase differences.

Slow Walking					
	ρ	RMSD	rRMSD	M	P
	Corr	RMSE	rRMSE	M	P
Subtalar Eversion	0.81	10.1 (3.5)	32.9 (9.6)	29.5 (36.3)	17.6 (10.1)
Ankle Plantarflexion	0.96	3.9 (1.2)	13.7 (4.0)	5.1 (14.0)	9.5 (3.4)
Knee Flexion	0.99	4.1 (2.4)	7.0 (4.3)	-0.3 (7.5)	4.7 (3.6)
Hip Abduction	0.91	4.1 (2.1)	27.6 (12.4)	-3.3 (42.8)	23.2 (10.9)
Hip External Rotation	0.76	6.7 (3.1)	39.5 (17.9)	12.9 (46.9)	13.0 (6.7)
Hip Flexion	0.99	5.2 (1.9)	13.3 (5.7)	-3.8 (16.4)	8.6 (4.2)

Figure 5: Slow walking speed; ankle, knee, and hip joint angle estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

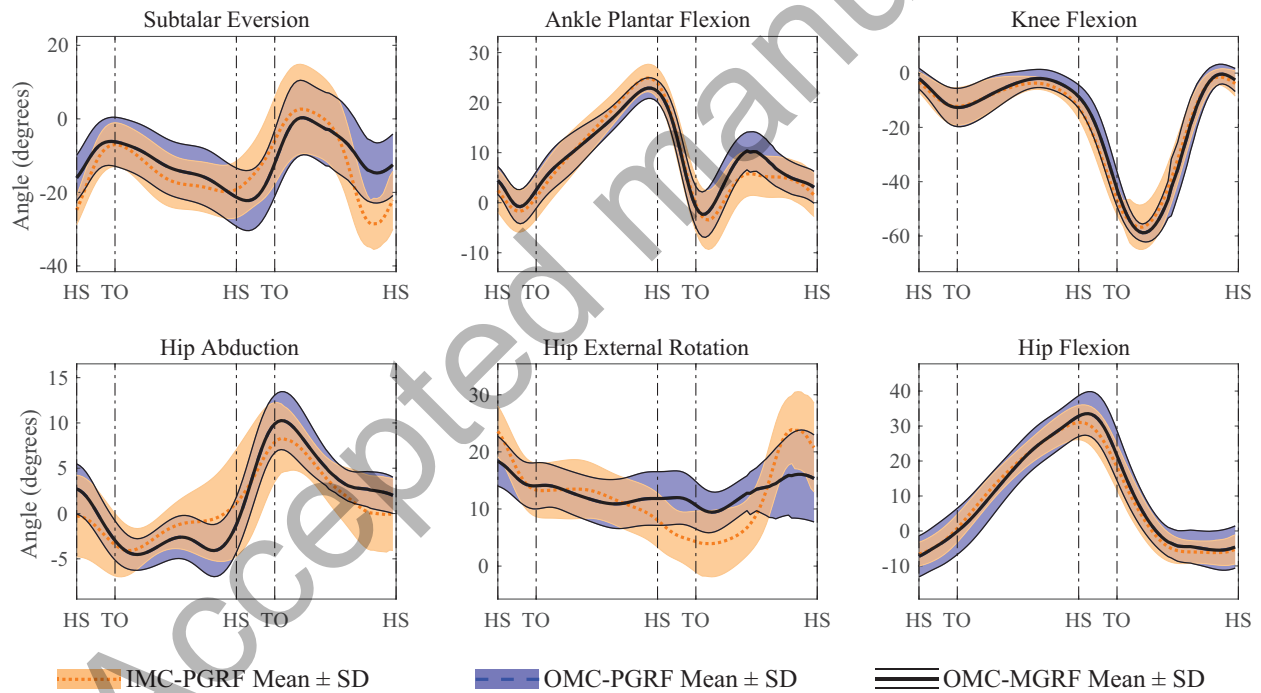


Table 6: Slow walking speed; IMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with *RMSD* (%BW or %BW*BH) and *rRMSD* (%), respectively. *M* and *P* indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.88	4.7 (0.8)	16.3 (2.3)	-26.3 (8.6)	15.8 (3.4)
Mediolateral	0.84	1.7 (0.5)	17.6 (3.6)	13.2 (23.1)	13.3 (3.5)
Vertical	0.97	8.1 (2.3)	7.4 (2.1)	-1.5 (1.2)	3.0 (0.9)
Frontal	0.64	1.0 (0.8)	45.7 (32.5)	177.8 (340.9)	32.6 (21.9)
Sagittal	0.90	1.5 (0.7)	18.7 (8.2)	10.2 (20.2)	12.2 (4.9)
Transverse	0.81	0.2 (0.1)	23.3 (5.8)	-0.7 (56.1)	17.5 (4.7)
Ankle					
Anteroposterior	0.85	22.8 (12.2)	29.0 (12.1)	56.5 (52.1)	10.7 (2.5)
Mediolateral	0.93	24.2 (11.3)	16.0 (6.7)	12.9 (21.3)	7.7 (2.9)
Proximodistal	0.93	87.4 (38.0)	14.3 (5.6)	8.2 (18.1)	7.1 (2.3)
Eversion	0.78	0.6 (0.3)	37.6 (27.2)	140.7 (259.7)	18.7 (13.5)
Plantar Flexion	0.93	1.6 (0.8)	16.8 (8.5)	9.6 (24.0)	10.1 (4.1)
Axial	0.65	0.5 (0.2)	33.7 (14.9)	47.8 (61.5)	27.9 (15.2)
Knee					
Anteroposterior	0.84	23.6 (6.8)	25.9 (11.3)	50.0 (65.3)	11.5 (3.0)
Mediolateral	0.89	10.5 (3.1)	14.7 (4.2)	3.8 (7.4)	7.3 (2.1)
Proximodistal	0.87	65.0 (29.7)	16.8 (7.3)	5.2 (11.7)	8.1 (3.1)
Abduction	0.74	1.2 (0.5)	23.2 (7.5)	-5.8 (20.4)	12.0 (4.5)
Flexion	0.46	1.8 (0.5)	34.5 (7.2)	35.3 (59.5)	36.9 (10.3)
Axial	0.60	0.3 (0.1)	29.8 (10.7)	8.2 (32.3)	33.0 (14.6)
Hip					
Anteroposterior	0.55	17.3 (7.2)	32.5 (10.3)	11.0 (30.6)	34.7 (12.2)
Mediolateral	0.66	23.8 (11.1)	25.1 (7.4)	5.8 (15.0)	11.1 (4.4)
Proximodistal	0.71	88.4 (23.7)	23.1 (3.8)	17.8 (8.4)	9.3 (2.4)
Abduction	0.83	1.4 (0.7)	21.0 (6.9)	-1.1 (13.4)	14.0 (8.3)
Flexion	0.92	2.0 (0.4)	22.8 (3.5)	93.3 (23.2)	15.5 (4.1)
External Rotation	0.57	0.4 (0.2)	32.4 (7.0)	-0.2 (40.5)	24.8 (10.4)

Table 7: Slow walking speed; OMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with *RMSD* (%BW or %BW*BH) and *rRMSD* (%), respectively. *M* and *P* indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.97	3.2 (1.2)	8.8 (2.6)	16.5 (13.0)	8.5 (1.7)
Mediolateral	0.82	1.5 (0.3)	17.2 (2.8)	4.4 (10.9)	13.8 (4.3)
Vertical	0.99	5.1 (1.6)	4.7 (1.5)	-1.4 (1.4)	1.8 (0.6)
Frontal	0.70	0.6 (0.2)	29.7 (10.2)	83.9 (124.3)	22.3 (9.5)
Sagittal	0.94	1.1 (0.4)	13.4 (5.0)	5.0 (12.4)	9.7 (4.2)
Transverse	0.80	0.2 (0.1)	21.2 (6.9)	16.2 (24.5)	18.1 (7.9)
Ankle					
Anteroposterior	0.83	16.8 (6.8)	22.7 (6.2)	32.4 (31.0)	11.2 (2.8)
Mediolateral	0.96	14.4 (3.8)	10.6 (3.1)	1.9 (10.0)	5.6 (2.4)
Proximodistal	0.97	55.3 (16.7)	9.8 (3.5)	1.2 (8.9)	5.2 (2.3)
Eversion	0.79	0.4 (0.1)	24.7 (8.3)	43.2 (59.0)	18.0 (13.3)
Plantar Flexion	0.96	0.9 (0.4)	10.6 (4.4)	-0.9 (9.2)	7.2 (3.2)
Axial	0.68	0.4 (0.1)	27.3 (7.3)	37.2 (42.0)	25.6 (11.9)
Knee					
Anteroposterior	0.92	9.9 (3.5)	14.8 (5.7)	-11.4 (11.5)	7.6 (2.7)
Mediolateral	0.96	7.0 (2.4)	9.9 (3.5)	-7.3 (6.1)	4.6 (1.3)
Proximodistal	0.95	39.1 (14.7)	10.1 (3.6)	-6.8 (5.6)	5.0 (1.6)
Abduction	0.90	0.7 (0.1)	14.0 (2.9)	-5.3 (11.3)	7.7 (1.6)
Flexion	0.84	0.8 (0.2)	18.7 (5.4)	-2.8 (18.0)	18.2 (5.8)
Axial	0.73	0.2 (0.1)	21.7 (7.0)	2.7 (16.2)	24.7 (9.2)
Hip					
Anteroposterior	0.87	8.3 (2.7)	17.0 (7.4)	-10.0 (9.0)	18.4 (9.3)
Mediolateral	0.90	12.9 (4.3)	13.7 (3.9)	-2.3 (5.7)	6.5 (2.0)
Proximodistal	0.91	44.3 (13.9)	12.4 (3.0)	-6.0 (5.9)	5.5 (1.3)
Abduction	0.93	0.7 (0.1)	12.4 (2.7)	-2.4 (6.2)	7.1 (1.7)
Flexion	0.81	1.1 (0.3)	19.2 (2.4)	-12.4 (11.7)	20.7 (3.3)
External Rotation	0.71	0.3 (0.1)	22.4 (4.1)	12.5 (16.0)	16.9 (4.2)

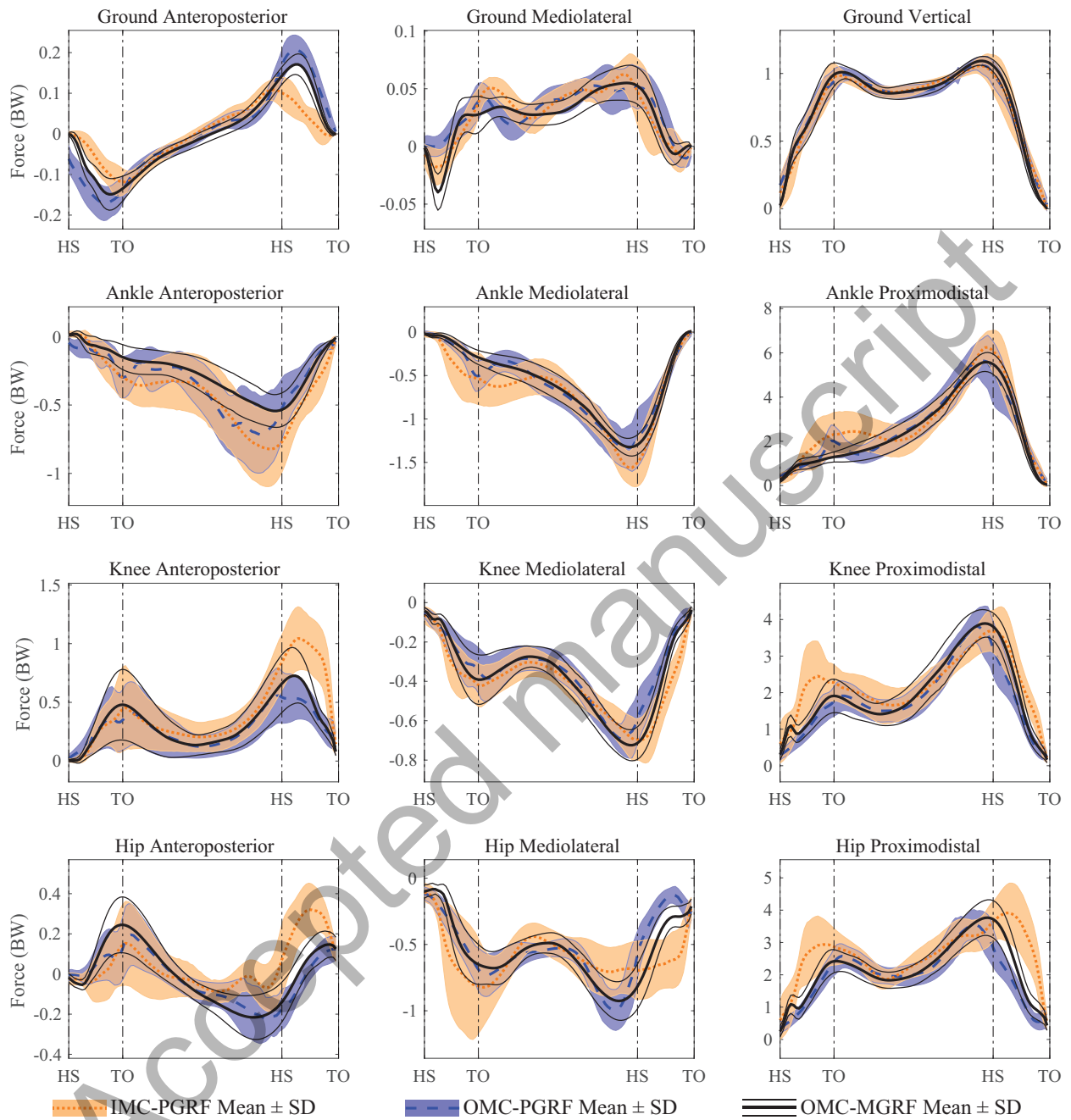


Figure 6: Slow walking speed; ground and lower limb joint reaction force estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

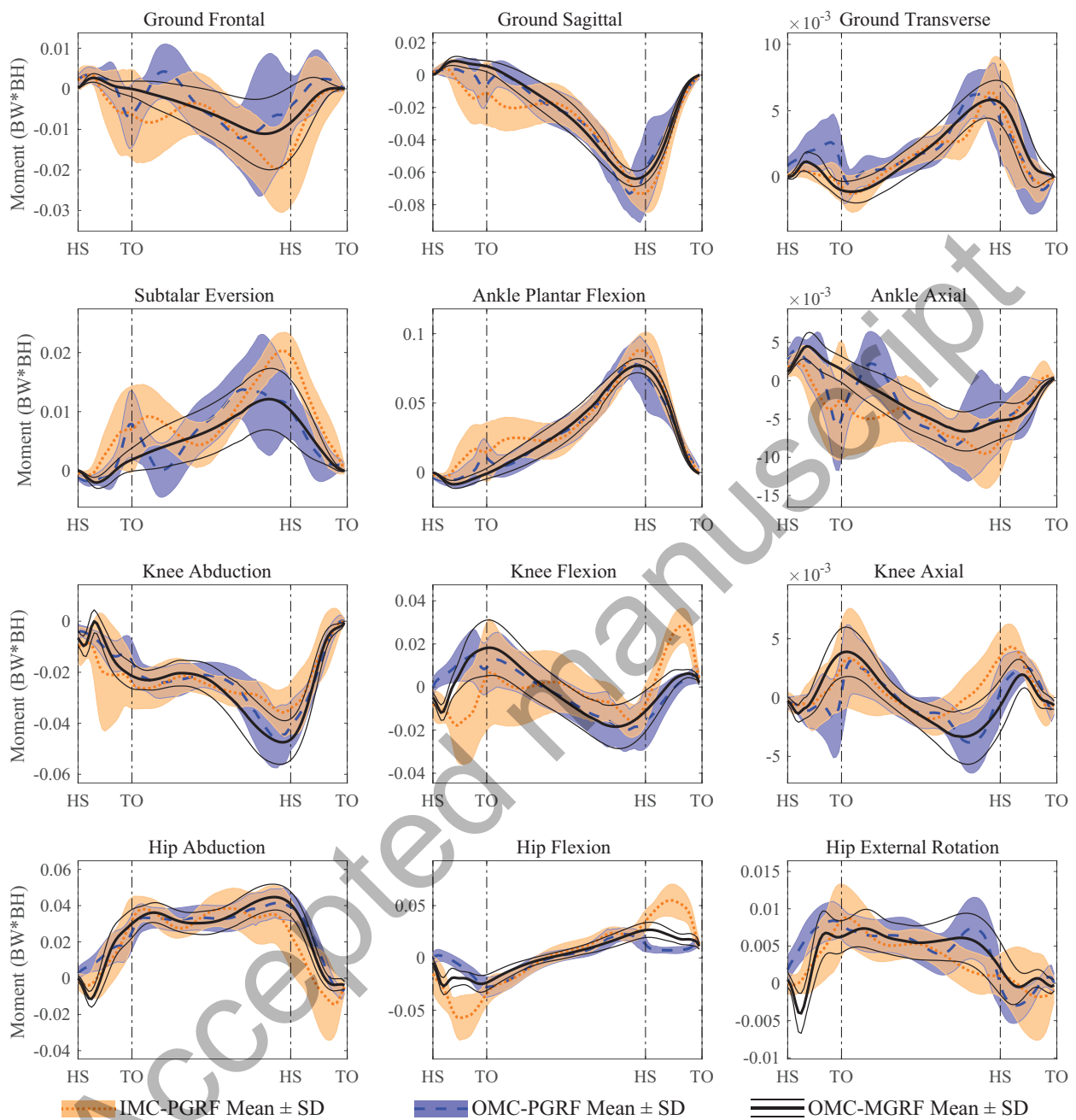


Figure 7: Slow walking speed; ground reaction and lower limb net joint moment estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

2.3. Fast walking speed

Table 8: Fast walking speed; comparison of lower limb joint angles between musculoskeletal model driven by the inertial (IMC-PGRF) and optical motion capture (OMC-PGRF/OMC-MGRF), using Pearson correlation coefficient (ρ), absolute and relative root-mean-squared-differences ($RMSD$ in degrees and $rRMSD$ in %, respectively). M and P denote the % magnitude and phase differences.

Fast Walking					
	ρ	$RMSD$	$rRMSD$	M	P
Subtalar Eversion	0.83	9.3 (2.8)	32.2 (11.4)	16.1 (29.9)	21.6 (10.9)
Ankle Plantarflexion	0.95	4.6 (1.3)	15.4 (5.1)	10.4 (18.3)	10.8 (4.7)
Knee Flexion	0.98	4.6 (1.6)	7.3 (2.6)	-0.0 (5.0)	4.6 (1.9)
Hip Abduction	0.9	4.2 (2.0)	25.0 (10.3)	-17.6 (29.6)	20.0 (8.4)
Hip External Rotation	0.62	6.2 (2.6)	35.3 (12.8)	-5.4 (29.1)	12.0 (6.4)
Hip Flexion	0.99	6.3 (1.9)	12.5 (4.4)	-4.6 (12.0)	9.1 (3.8)

Figure 8: Fast walking speed; ankle, knee, and hip joint angle estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

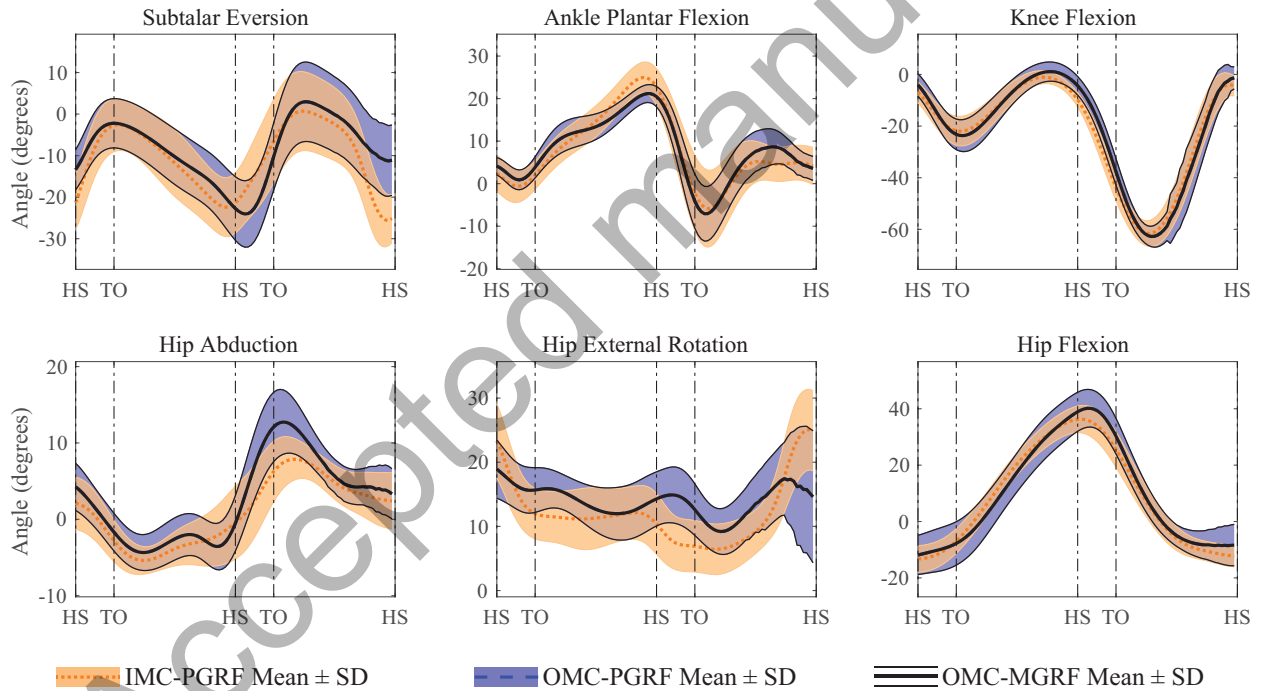


Table 9: Fast walking speed; IMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with *RMSD* (%BW or %BW*BH) and *rRMSD* (%), respectively. *M* and *P* indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.92	6.5 (1.2)	14.1 (2.0)	-25.2 (6.6)	13.0 (2.7)
Mediolateral	0.75	2.5 (0.7)	19.3 (3.4)	3.9 (15.2)	17.5 (3.8)
Vertical	0.95	11.5 (3.2)	8.8 (2.2)	-1.7 (1.8)	4.1 (1.1)
Frontal	0.61	0.9 (0.5)	34.1 (15.5)	91.7 (269.9)	30.9 (15.0)
Sagittal	0.90	1.7 (0.4)	17.1 (4.2)	21.6 (15.2)	12.7 (4.1)
Transverse	0.81	0.2 (0.1)	24.4 (9.3)	-13.1 (33.0)	18.4 (6.8)
Ankle					
Anteroposterior	0.84	22.0 (8.0)	23.4 (6.9)	41.8 (34.5)	10.7 (1.5)
Mediolateral	0.94	25.0 (6.2)	14.9 (3.7)	17.7 (13.1)	8.0 (2.9)
Proximodistal	0.93	93.3 (23.8)	13.6 (3.2)	13.8 (9.3)	7.5 (2.5)
Eversion	0.75	0.7 (0.2)	30.7 (13.7)	81.9 (137.1)	19.8 (9.8)
Plantar Flexion	0.93	1.6 (0.4)	14.1 (4.2)	14.8 (12.5)	10.1 (3.3)
Axial	0.65	0.6 (0.1)	28.3 (8.4)	54.9 (33.2)	28.6 (12.1)
Knee					
Anteroposterior	0.75	39.4 (8.2)	25.3 (7.4)	29.3 (38.3)	15.7 (4.6)
Mediolateral	0.90	14.5 (3.0)	14.3 (3.3)	8.0 (8.9)	7.3 (1.5)
Proximodistal	0.92	66.8 (19.0)	12.8 (4.0)	5.0 (7.0)	6.7 (1.8)
Abduction	0.86	1.1 (0.2)	15.5 (3.3)	0.1 (10.9)	9.8 (2.1)
Flexion	0.67	2.0 (0.4)	24.6 (5.3)	-0.2 (18.7)	28.2 (8.3)
Axial	0.83	0.3 (0.1)	20.6 (8.1)	-4.9 (25.8)	22.2 (11.5)
Hip					
Anteroposterior	0.80	19.0 (8.3)	22.5 (6.7)	3.3 (23.4)	20.7 (5.9)
Mediolateral	0.77	31.8 (11.6)	21.1 (6.8)	10.5 (13.0)	10.6 (3.7)
Proximodistal	0.82	123.9 (32.5)	20.3 (4.9)	21.7 (11.2)	9.2 (2.4)
Abduction	0.80	1.5 (0.6)	19.5 (4.5)	10.7 (16.6)	14.4 (6.9)
Flexion	0.91	2.6 (0.6)	16.4 (3.1)	55.2 (20.0)	14.5 (3.8)
External Rotation	0.44	0.5 (0.2)	31.7 (5.6)	-4.5 (36.2)	26.6 (9.9)

Table 10: Fast walking speed; OMC-PGRF-based ground and joint reaction forces (first three quantities) and net moments (second three quantities) versus OMC-MGRF. Pearson correlation coefficient is denoted with ρ . Absolute per body weight (or body weight times height) and relative root-mean-squared-difference are denoted with $RMSD$ (%BW or %BW*BH) and $rRMSD$ (%), respectively. M and P indicate the magnitude and phase differences (%).

	ρ	RMSD	rRMSD	M	P
Ground					
Anteroposterior	0.96	4.2 (0.8)	7.9 (1.5)	0.1 (7.4)	9.0 (1.7)
Mediolateral	0.76	2.2 (0.5)	19.4 (5.3)	1.2 (10.6)	16.5 (6.0)
Vertical	0.98	7.2 (2.3)	5.4 (1.6)	-1.0 (0.9)	2.6 (0.8)
Frontal	0.59	0.8 (0.2)	30.1 (8.1)	59.1 (92.8)	26.9 (9.3)
Sagittal	0.94	1.5 (0.3)	14.3 (2.9)	28.3 (13.3)	9.7 (2.8)
Transverse	0.80	0.2 (0.1)	21.0 (8.5)	1.1 (19.8)	17.9 (8.3)
Ankle					
Anteroposterior	0.83	21.3 (7.0)	22.9 (5.2)	39.4 (23.6)	11.0 (1.9)
Mediolateral	0.95	19.4 (4.2)	11.9 (2.3)	12.5 (8.6)	6.6 (2.2)
Proximodistal	0.96	76.4 (15.8)	11.0 (2.0)	13.7 (6.9)	5.7 (1.7)
Eversion	0.72	0.6 (0.1)	26.2 (5.6)	47.5 (55.2)	20.3 (7.5)
Plantar Flexion	0.96	1.3 (0.3)	10.9 (2.6)	13.8 (8.0)	7.7 (2.7)
Axial	0.60	0.5 (0.1)	27.0 (8.2)	33.3 (32.7)	30.6 (12.1)
Knee					
Anteroposterior	0.94	15.2 (4.1)	10.8 (2.3)	-3.0 (5.8)	7.4 (1.6)
Mediolateral	0.96	8.2 (1.6)	8.5 (1.4)	-0.6 (4.7)	4.5 (0.8)
Proximodistal	0.95	49.6 (9.0)	9.4 (1.6)	1.7 (4.7)	5.3 (0.9)
Abduction	0.92	0.9 (0.2)	11.8 (1.8)	6.1 (9.0)	7.6 (1.5)
Flexion	0.90	1.2 (0.3)	14.6 (3.4)	-1.1 (11.2)	15.2 (4.0)
Axial	0.89	0.2 (0.1)	14.8 (4.7)	-6.5 (18.5)	16.1 (4.9)
Hip					
Anteroposterior	0.91	11.9 (4.0)	15.4 (6.4)	-9.1 (13.3)	14.4 (5.4)
Mediolateral	0.94	16.6 (3.5)	11.7 (2.4)	-1.2 (7.9)	5.8 (1.2)
Proximodistal	0.93	59.5 (17.0)	10.8 (2.5)	-2.6 (7.2)	5.5 (1.3)
Abduction	0.88	1.0 (0.2)	14.3 (2.5)	-3.5 (7.2)	10.4 (2.4)
Flexion	0.89	1.6 (0.3)	13.8 (2.9)	-6.1 (13.0)	15.3 (3.8)
External Rotation	0.67	0.4 (0.1)	22.2 (3.5)	-0.9 (12.6)	20.7 (5.3)

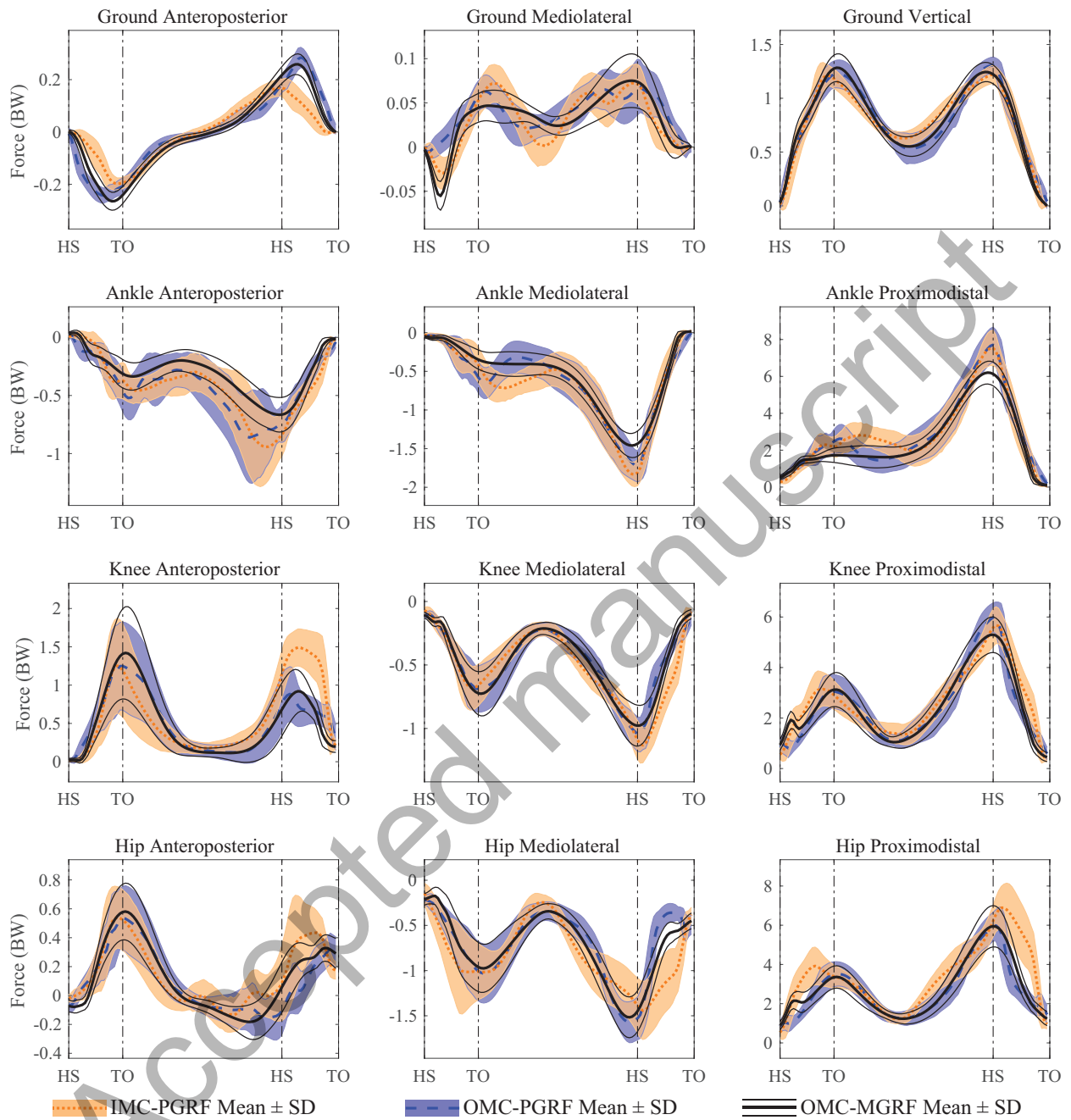


Figure 9: Fast walking speed; ground and lower limb joint reaction force estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid lines around thick black solid line).

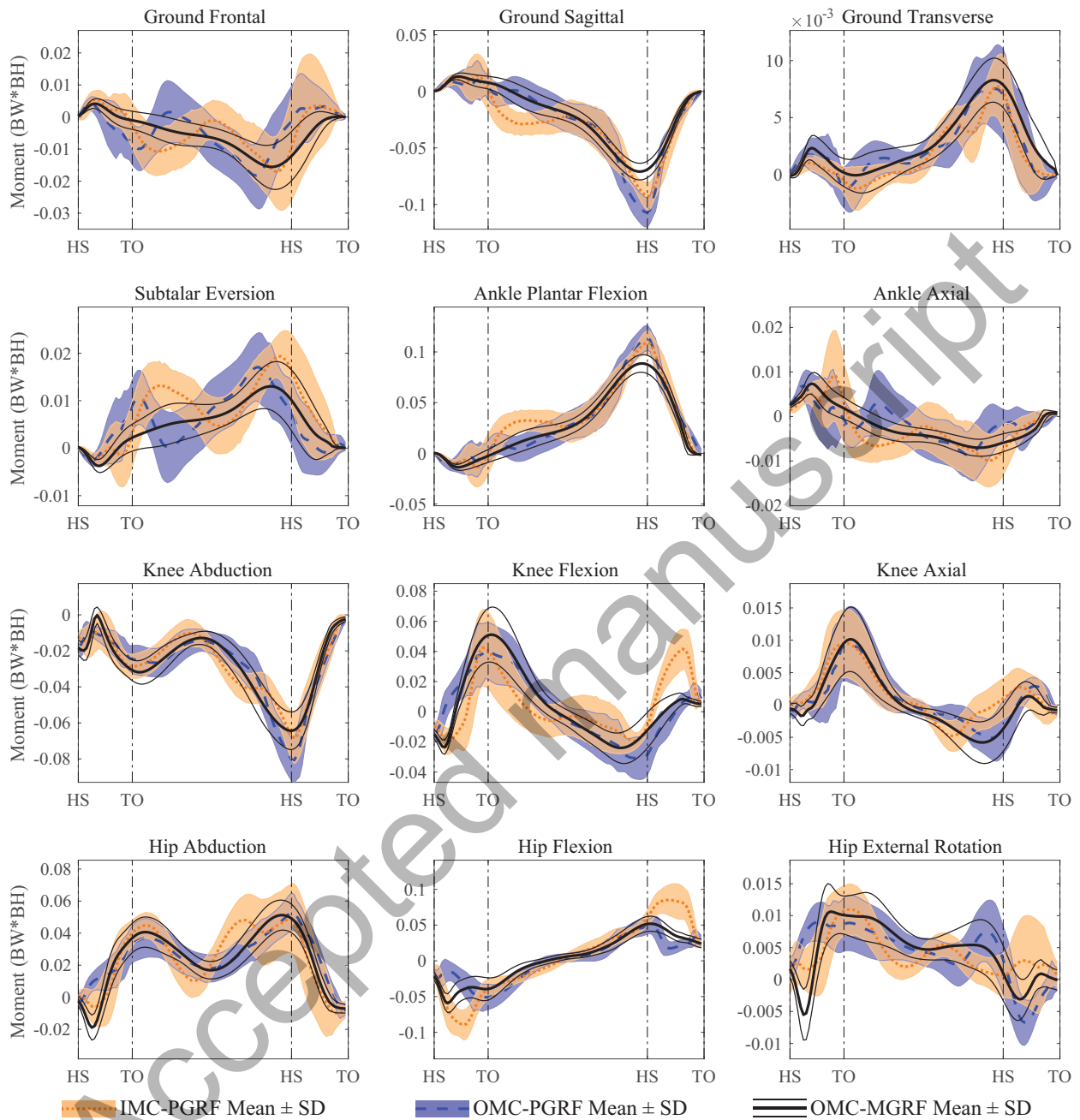


Figure 10: Fast walking speed; ground reaction and lower limb net joint moment estimates (standard deviation around mean) of the IMC-PGRF (orange shaded area around orange dotted line) and OMC-PGRF models (blue shaded area around blue dashed line) versus OMC-MGRF model (thin black solid line around thick black solid line).