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Gait analysis of transfemoral amputees: errors in inverse dynamics are substantial and depend on prosthetic design

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Abstract
Quantitative assessments of prostheses performances rely more and more frequently on gait analysis focusing on prosthetic knee joint forces and moments computed by inverse dynamics. However, this method is prone to errors, as demonstrated in comparison with direct measurements of these forces and moments. The magnitude of errors reported in the literature seems to vary depending on prosthetic components. Therefore, the purposes of this study were (A) to quantify and compare the magnitude of errors in knee joint forces and moments obtained with inverse dynamics and direct measurements on ten participants with transfemoral amputation during walking and (B) to investigate if these errors can be characterised for different prosthetic knees. Knee joint forces and moments computed by inverse dynamics presented substantial errors, especially during the swing phase of gait. Indeed, the median errors in percentage of the moment magnitude were 4% and 26% in extension/flexion, 6% and 19% in adduction/abduction as well as 14% and 27% in internal/external rotation during stance and swing phase, respectively. Moreover, errors varied depending on the prosthetic limb fitted with mechanical or microprocessor-controlled knees. This study confirmed that inverse dynamics should be used cautiously while performing gait analysis of amputees. Alternatively, direct measurements of joint forces and moments could be relevant for mechanical characterising of components
and alignments of prosthetic limbs.

Keywords
Above-knee amputation, artificial limb, bone-anchored prosthesis, joint moments, kinetics, loading, load cell, osseointegration, transducer, validation

I-INTRODUCTION

Clinical examinations leading to objective evaluations of ambulation abilities of individuals with lower-limb amputation are increasingly required. Typically, quantitative assessments of prostheses performances rely on spatiotemporal, kinematic and kinetic gait characteristics [1]–[6].

In particular, the analysis of lower limb joints kinetics (i.e., forces, moments, power) has become critical to compare mechanical performances between adaptive dissipation prosthetic knee units [7]–[19] and an anatomical knee joint [2], [3], [20]. Furthermore, the development of osseointegrated fixations for bone-anchored prostheses requires a better understanding and monitoring of implant and prosthetic loading during locomotion to increase walking abilities (e.g., speed of walking) while assuring safety (e.g., limitation of high loading, fall prevention, breakage of fixation parts) [6], [21]–[29].

One way to produce such knee joint kinetics is to rely on inverse dynamics computations. Unfortunately, joint forces and moments obtained this way tend to be prone to errors especially for prosthetic gait [30]–[35]. These errors could be mainly attributed to inaccurate measurements of prostheses inertial parameters and oversimplified modelling of prosthetic segments (i.e., rigid) and prosthetic joints (i.e., with constant centre/axis of rotation, without any damping nor friction).

However, prosthetic gait provides a singular opportunity to validate the computation of knee joint kinetics by comparing knee forces and moments obtained with inverse dynamics equations with the ones measured directly by a transducer fitted within the prosthesis [36]–[40].

Previous studies comparing both methods involving participants fitted with various types of knees revealed errors close to 5% of body weight and 30% of body weight times height for knee joint forces and moments, respectively [38]. Interestingly, the magnitude of errors seems to vary between these studies involving various prosthetic components (e.g., one participant with a constant friction knee [36] versus six participants with hydraulic microprocessor-controlled knees [38]). One could hypothesize that the range of these errors could be attributed to differences in absorption in the foot and dissipation in knee components that are hardly taken into account in inverse dynamic computations while, conversely, properly assessed by direct measurements.

Clearly, there is a need for a more in-depth investigation of the magnitude of errors in joint forces and moments obtained with inverse dynamics during walking with various types of prosthetic components.

Therefore, this present study capitalized on unique kinematic and dynamics datasets initially collected for a study assessing walking abilities of participants with bone-anchored prostheses [6], [21], [24], [41]. Consequently, the purposes of this retrospective study were: (A) to quantify and compare the magnitude of errors in prosthetic knee joint forces and moments obtained with inverse dynamics and direct measurements for ten participants with unilateral transfemoral amputation fitted a bone-anchored prosthesis including different types of hydraulic knees during walking, and (B) to investigate if these types of knees (i.e., mechanical, microprocessor-controlled) could have an effect on these errors.

II-MATERIAL AND METHODS

A. Participants

The eleven participants initially recruited by Sahlgrenska University Hospital, Sweden represented approximately 15% of the population worldwide at the time of recording [6], [21], [24], [41]. One subject initially tested was discarded in this study because some measures required to calculate the inertial characteristics of the residuum were not recorded. Each participant was fully rehabilitated, fitted with the fixation for at least one year, able to walk 200 m independently, weighed less than 110 kg to avoid overloading the transducer and reported no incidents (e.g., falls) six months prior to the recording [25], [27], [28]. Human research ethical approval was received from the Queensland University of Technology (0600000451). Written consent was obtained from all participants.

All the participants were fitted with the OPRA implant system (Integrum, AB, Sweden). The seven males and three females were assessed during walking with their own prosthesis (i.e., knee, feet, footwear) to warrant ecological evaluations. Nonetheless, they all used hydraulic knee units including either a Total Knee 2000 or Mauch Knee (Ossur, Iceland) or a C-Leg (Ottobock, Germany). The participants' demographics are detailed in Table I.

*** Insert Table 1 ***

B. Measurements

Participants performed three successive trials of straight level walking at self-selected speed. Kinematic and overall dynamic data were recorded simultaneously with a 6-camera ProReflex 240 3D Motion Capture Unit (Qualisys, Gothenburg, Sweden) and two force plates (Kistler, Jonsered, Sweden), respectively. Relevant markers placed on anatomical landmarks on the pelvis, residuum, hydraulic knee, pylon and shoe were used (i.e., midpoint between posterior iliac spines, anterior superior iliac spines, greater trochanter, lateral epicondyle, tibial tuberosity, lateral malleolus, calcaneum, and fifth metatarsal head).

The residuum dynamic data were measured directly using a multi-axial transducer (JR3 Inc., Woodland, CA, USA) fitted between the osseointegrated fixation and the knee joint as described in previous publications [6], [18], [21], [22], [24], [25], [27]–[29], [42], [43]. The coordinate system of the transducer was manually aligned with the residuum and prosthetic knee anatomical axes. Markers were also placed on the front, back and side of the transducer to define its position in the inertial coordinate system (ICS). The forces and moments were recorded directly onto a laptop connected to the transducer via a serial cable using a customized LabView (National Instruments Corporation, USA) program. All data sets were recorded at 200 Hz.

The inertial parameters of the thigh (i.e., residuum, transducer, connecting pylons, prosthetic knee), shank (i.e., connecting pylons) and foot (i.e., prosthetic ankle, foot, and shoe) segments of the prosthetic limb were calculated using typical geometrical shapes of each component based on bench-top measurements on the component's medio-lateral, antero-posterior and long axes [44].

*** Insert Figure 1 ***

C. Data Processing

Individual heel contact and toe-off events were identified manually using the force applied on the superior/inferior axis provided by the transducer. All data sets were manually synchronized a posteriori, using the superior/inferior force during first heel contact on force-plate.

The kinematic, overall dynamic and residuum dynamic data were purposely smoothed with a basic Hanning's algorithm relying on sliding window of five samples with following coefficients: T − 2 = 0.15, T − 1 = 0.20, T = 0.30, T + 1 = 0.20, T + 2 = 0.15. This method was chosen based on the premises that more advanced filtering could potentially reduce differences in results of forces and moments.

The forces and moments were computed at the knee joint using 3D inverse dynamics [36], [45] and were expressed with respect to the thigh segment coordinate system. Knee kinetic data for amputation either on the left or right side were transformed from the transducer centre to the knee joint centre [36] and determined so that the forces were positive laterally, anteriorly and superiorly for the lateral/medial (Lat/Med), anterior/posterior (Ant/Post), and superior/inferior (Sup/Inf) directions while the moments were positive in extension, adduction and internally for the extension/flexion (Ext/Flex) adduction/abduction (Add/Abd), and internal/external (Int/Ext) rotations, respectively. The knee joint centre was defined at a constant distance of 8 cm from the lateral epicondyle marker along with the knee flexion axis. This point is not representative of the knee mechanics but a conventional point where to express the joint moment.

Then, some optimisations were done to compensate for potential errors due to the fitting of the transducer and the determinations of segmental inertial characteristics of the prosthetic limb. First, the position and orientation the transducer with respect to thigh segment origin and axes were refined by an optimisation, aiming at minimising the differences in forces and moments at the knee joint during the stance phase of the gait cycle, mainly influenced by the alignment of the transducer during this phase. The design variables were three translation components and three rotation angles defining the transformation from the transducer centre to the knee joint centre. The initial guess was the position of the midpoint between the two markers placed on the side of the transducer and three zeros. The objective function was the unweighted sum over all the frames of the stance phase of the squared errors between the measured (and transformed) and the computed three force and moment components expressed in the thigh segment coordinate system. The solution was obtained by a Quasi-Newtow algorithm (function fminunc in Matlab). Second, the inertial parameters were refined using an optimisation aiming at minimising the same differences but during the swing phase of the gait cycle, mainly influenced by the inertial parameters during this phase. The design variables were the mass, position of centre of mass and moments of inertia of the foot and shank segments. The initial guess was the inertial parameters obtained by the bench-top measurements. The objective function was the same as the first optimisation but with the sum over all the frames of the swing phase. The solution was obtained by a constrained optimisation algorithm (function fmincon in Matlab) limiting the solution in an interval of ±15% of the initial guess. Finally, all gait cycles were resampled on 100 points, representing 100% of the gait cycle.

The knee joint forces and moments computed by each method using the adjusted transducer's position and orientation and inertial parameters were compared. The comparison involved calculation of the root mean square errors (RMSE) for the three force and moment components in the thigh segment coordinate system. RMSE were computed for each participant considering the three gait trials collated together (i.e., 300 points). The errors were expressed as a percentage of the measured amplitudes (RMSE%) separating the stance and swing phases. The RMSE% were characterised by the median, lower and upper quartiles (interquartile range: IQR), minimum and maximum for the participants (n = 10, three gait trials collated) as well as for the different hydraulic knees (n = 6 for Total Knee, n = 2 for Mauch Knee, and n = 2 for C-Leg, three gait trials collated). Because of small sample size, the statistical effect of the hydraulic knee was tested with a permutation test (i.e., independent sample permutation-based t-test with α-level of 0.05) [46].

III. RESULTS

Within the two optimisations, the median and IQR RMSE for the whole gait cycle were modified from 57.3 N (45.4–110.4) to 29.5 N (17.4–34.7) and to 23.8 N (9.9–33.3) for the Ant/Post force; from 32.5 N (28.5–53.4) to 32.0 N (25.9–49.9) and to 24.5 N (12.3–41.1) for the Sup/Inf force; from 61.4 N (38.0–93.0) to 8.4 N (5.8–12.0) and to 6.5 N (3.8–10.6) for the Lat/Med force; from 16.2 N.m (13.0–24.4) to 4.1 N.m (3.4–9.6) and to 3.9 N.m (1.8–7.8) for the Ext/Flex moment; from 26.9 N.m (19.1–29.4) to 2.5 N.m (1.6–4.5) and to 1.7 N.m (0.8–3.4) for the Add/Abd moment; and from 4.0 N.m (3.5–7.6) to 1.6 N.m (1.2–2.8) and to 1.2 N.m (0.6–2.3) for the Int/Ext moment.

All trials considered, the RMSE% were

generally lower during stance than swing, almost of the same for forces and moments during stance and higher for joint forces than for joint moments during swing (Fig. 2).

*** Insert Figure 2 ***

The statistical comparison of the three hydraulic knees (Fig. 3) indicated four significant differences for the joint moment during swing. For the Ext/Flex moment during swing, the RSME% of 16 (14–24) for Total Knee were significantly lower compared to 35 (32–38) for Mauch Knee ($p = 0.0356$) and 41 (37-46) for C-Leg ($p = 0.0330$). Similarly, for the Add/Abd moment during swing, the RSME% of 16 (15–22) for Total Knee were significantly lower compared to 29 (28–30) for Mauch Knee ($p = 0.0318$) and 19 (18–20) for C-Leg ($p = 0.0344$). For the Int/Ext moment during swing, the RSME% of 25 (20–27) for Total Knee were lower than 39 (29–48) for C-Leg but not significantly ($p = 0.0874$). The results for the joint moments during stance and for the joint forces during both stance and swing reveal no significant differences. These results are provided as supplementary figures.

Typical results for the three gait cycles of three participants with different hydraulic knees are given in Fig. 4. The patterns of joint forces and moments were similar. However, Ant/Post force and Ext/Flex moment presented some differences during the swing phase. For instance, the Ext/Flex moment computed by inverse dynamics was comparable to the one measured directly by the transducer outside the instantaneous spikes for participant #6 (Total Knee). Conversely, the Ext/Flex moment appeared underestimated for participants #8 (Mauch Knee) and #5 (C-Leg).

*** Insert Figure 3 ***

IV. DISCUSSION

The patterns and magnitudes of the knee joint forces and moments computed by the inverse dynamics and directly measured were typical of individuals with a transfemoral amputation during walking [10], [36], [38], [47], [48].

The ranges of errors were also consistent with previous studies [36], [38]. The average errors reported for Ext/Flex, Add/Abd and Int/Ext knee joint moments obtained with a transducer placed below the knee of individuals with a transtibial amputation were 12%, 11% and 22% of the moment amplitude during the swing phase [40] while, in the present study, the median RMSE% were 26, 19, and 27, respectively. This suggests that some of the errors may be due to the mechanics of the knee prosthesis and mainly due to the Ext/Flex hydraulic control. Indeed, in the inverse dynamics, the knee joint is assumed to have a constant axis of rotation and no damping nor friction. Therefore, some differences in RMSE% should fairly be observed between mechanical and microprocessor controlled hydraulic knees. In the present study, the observed differences in Ext/Flex and Add/Abd moments during swing are a lower magnitude of errors for the Total Knee (except for errors due to some spikes at the end of the swing phase) and a higher level of errors for Mauch Knee and for C-Leg. These differences were found statistically significant according to the permutation-based t-test.

The outcomes of this study were limited by the number of participants as often in prosthetic research involving cumbersome experimental protocol (e.g., multi-axial transducer, motion analysis system and force plates altogether). For instance, previous studies comparing inverse dynamics and direct measurements involved between one to seven participants [36]–[40]. Incidentally, the group tested here represented approximately 15% of existing population of individuals fitted with an osseointegrated implant worldwide at the time of recording. Other limitations were due to the inverse dynamics and direct measurement methods: manual synchronisation, estimation of the residuum and prosthesis inertial parameters, position and orientation of the transducer.

The differences in the RMSE% in both knee joint forces and moments due to $+/-1$ frame (at 200 Hz) were estimated lower than $+/-1\%$ and 2% on the median and IQR during stance and lower than $+/-3\%$ and 4% during swing. Furthermore, two optimisations were successively performed to minimise the effects of potential errors in inertial parameters and transducer's setting. Nonetheless, more definitive evidence of the differences between inverse dynamics and direct measurement methods, particularly during the swing phase, would require relying on actual initial characteristics of the prostheses obtained with the pendulum method. Consequently, altogether, the generalisation of the results presented here must be conducted carefully.

However, the present study seemed to indicate that the joint forces and moments computed by inverse dynamics could present substantial errors. It can be understood that the computation of the segment accelerations by time derivation of markers trajectories could lack accuracy to reflect the damping effects of the prosthetic components, and in particular the hydraulic control of the different prosthetic knees. Therefore, the range of the errors seems consistent with the knee prosthetic designs: the more hydraulic control, the larger the errors.

Indeed, the Total Knee is a polycentric knee with a 3-phase swing control, the Mauch Knee is a single axis knee with distinct stance and swing control and C-Leg is a single axis knee with microprocessor control. Moreover, errors due to spikes at the end of the swing phase were also observed with a constant friction knee [36]. Alike the knee dissipation hardly assessed by inverse dynamics, these spikes are representative of the prosthetic knee mechanics, namely the knee unlocking mechanism and the limit stop at the ends of the stance and swing phases of gait, respectively [42].

*** Insert Figure 4 ***

V. CONCLUSION

Inverse dynamics might be used with caution in prosthetics given the magnitude of the errors in the joint forces and moments suggested in this study. Accelerometers or gyroscopes based methods could be helpful to better compute the joint forces and moments in persons with lower limb amputation [49], [50]. Alternatively, direct measurements of loading could be relevant and reliable for the mechanical characterisation of components and alignments of prosthetic limbs [51], [52].

VI-TO KNOW MORE

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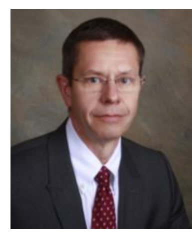

Rickard Brånemark was born in Malmö, Sweden, in 1960. He received the M.D. degree from Göteborg University, Sweden, in 1987, the M.Sc. degree in technical physics from Chalmers University of Technology, Göteborg, Sweden, in 1987, and the Ph.D. degree from Department of Orthopedics, Institute of Surgical Sciences and Institute of Anatomy and Cell Biology, Göteborg University, Sweden, in 1996.

He was appointed Associate Professor at the Department of Orthopaedics, at University of Gothenburg in 2012. He is presently visiting associate professor at the Department of Ortopaedic Surgery at University of California, San Francisco. Dr. Brånemark is the President of the Orthopaedic Surgical Osseointegration Society; he is a member of the Swedish Orthopaedic Association, the Swedish Society of Medicine, the International Society for Prosthetics and Orthotics, the American Association of Orthopaedic Surgeons and the Royal Society of Medicine. He has pioneered the studies on titanium implants to anchor limb prostheses. He has been internationally acknowledged, and he was rewarded with the Hanger Prize, 2007, for his contribution in amputation research.

Laurent Frossard was born in Nevers, France, in 1970. He received the PhD degree in fundamental and applied sciences, biomechanics from University of Poitiers, France, in 1998. He has held a number of teaching and research positions in academia and private sectors in France, Canada, and Australia (www.LaurentFrossard.com).

He is currently an Adjunct Professor at the Queensland University of Technology and the University of the Sunshine Coast in Australia. He is an active Researcher focusing on the developments of biomechanical tools and basic knowledge of the locomotion of individuals with lower limb loss during rehabilitation and activities of daily living. He is the leader and manager of several large-scale projects founded by competitive grants in partnerships with multiple commercial entities in Australia and overseas.

He is also an educator (e.g., biomechanics, research methods, research project management) and an entrepreneur (www.YourResearchProject.com). He has authored over 130 publications (i.e., articles, conference papers, book chapters and books) leading to international recognition as an independent expert on the clinical outcomes of bone-anchored prostheses.

FIGURE 1. Overview of conventional calculation and direct measurement of knee joint forces and moments for individuals with transfemoral amputation highlighting the computing with inverse dynamics and directmethods as well as the recording of input data including the setup of motion analysis laboratory, market set and prosthetic attachment featuring the residuum (A), osseointegrated implant (B), pyramidal adaptors (C), serial cable (D), transducer (E), and knee joint (F). DoW: Direction of Walking, ICS: Inertial Coordinate System, SCS: thigh Segment Coordinate System, Lat/Med: lateral/medial axis, Ant/Post: anterior/posterior axis, Sup/Inf: superior/inferior axis.

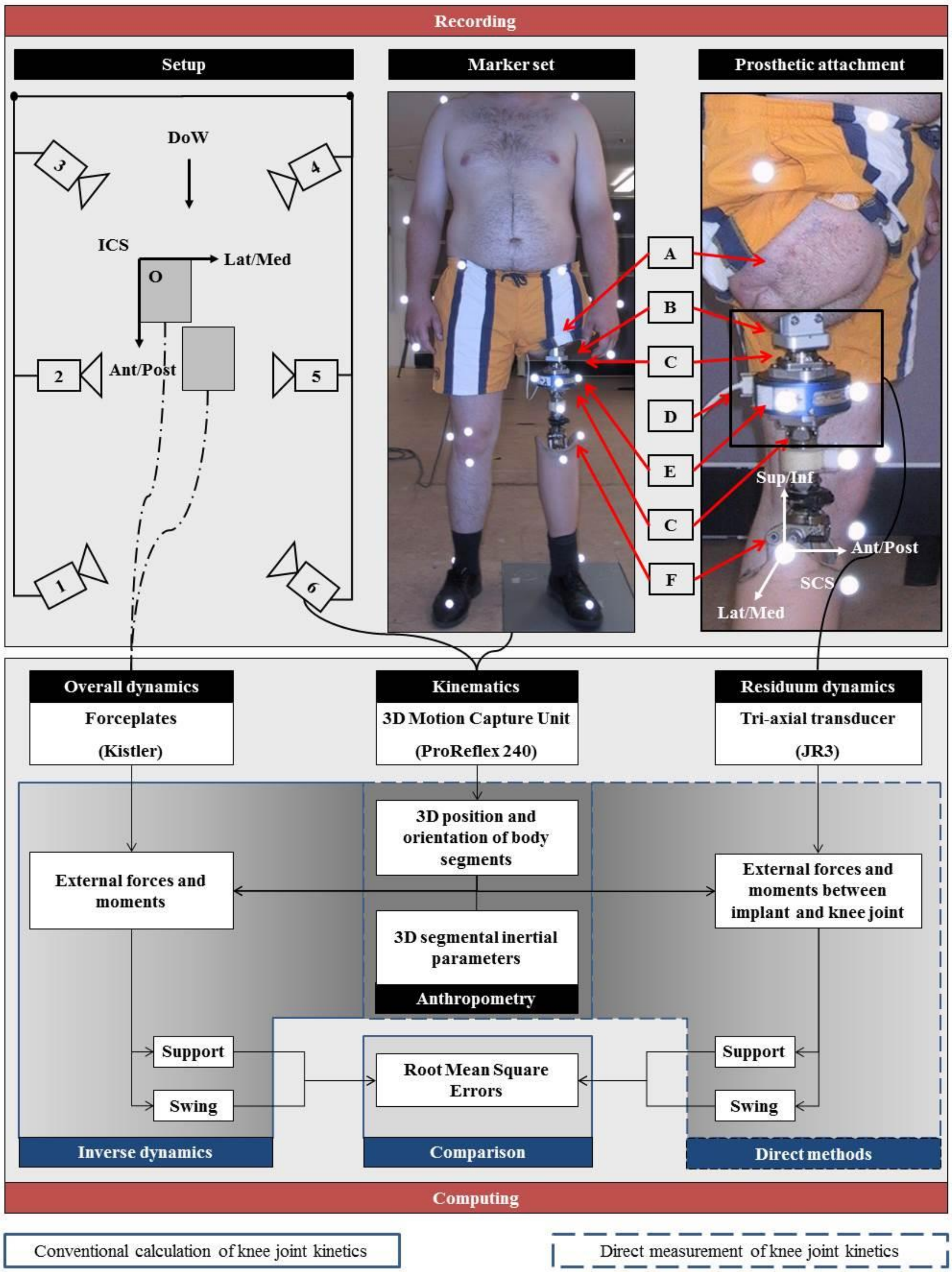

FIGURE 2. Box-and-Whisker plots (i.e., median, lower and upper quartiles, minimum and maximum, and outlier "+") of the RMSE% for the ten participants (n = 10, the three gait trials of each participant collated together) during the stance and swing phases of gait.

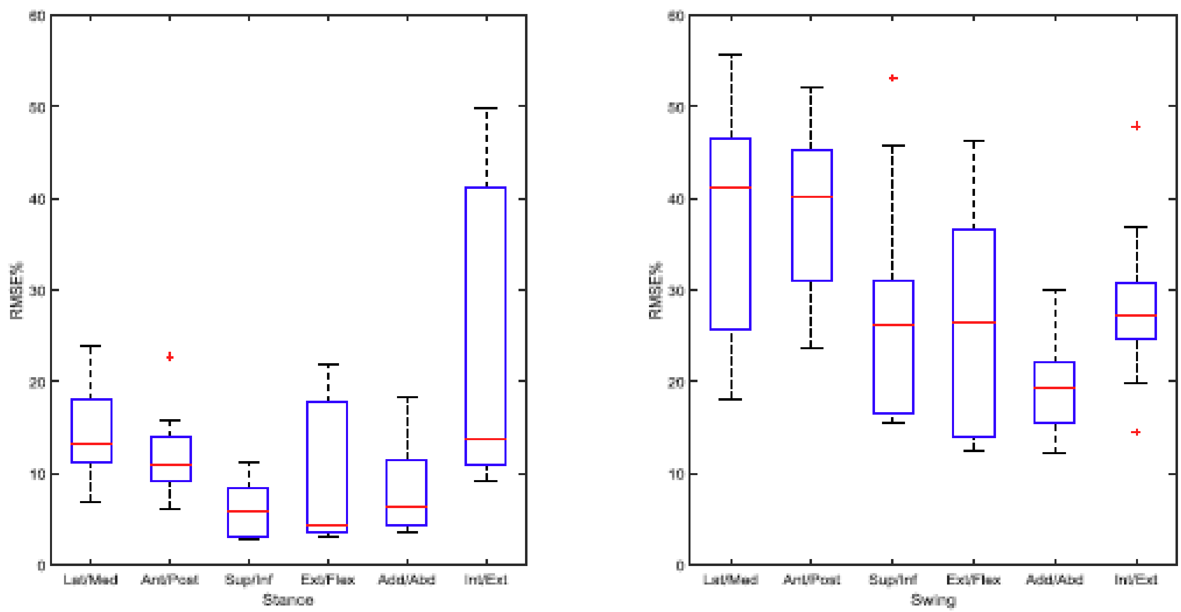

FIGURE 3. Box-and-Whisker plots (i.e., median, lower and upper quartiles, minimum and maximum, and outlier "+") of RMSE% for extension/flexion (Ext/Flex), adduction/abduction (Add/Abd) and internal/external rotation (Int/Ext) moments for the three hydraulic knees (n = 6 for Total Knee, n = 2 for Mauch Knee, and n = 2 for C-Leg, the three gait trials of each participant collated together) during the swing phase of gait.

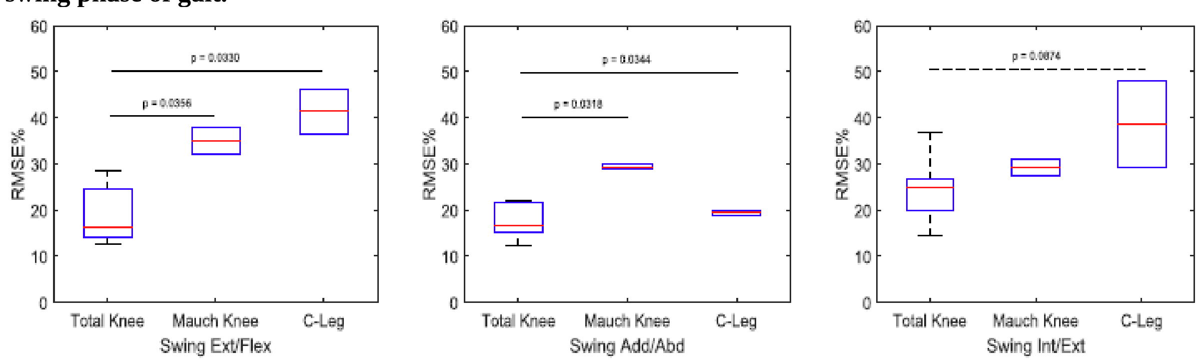

FIGURE 4. Examples of joint forces and moments for the three gait trials superimposed of participants #6 (Total Knee), #8 (Mauch Knee), and #5 (C-Leg), from top down, computed by inverse dynamics (in blue) and measured by the transducer (in red). The reader may refer to the online version of the article for coloured figure. Dashed vertical line indicates toe-off. RMSE% (the three gait trials of each participant collated together) are given for the stance and swing phases of gait.

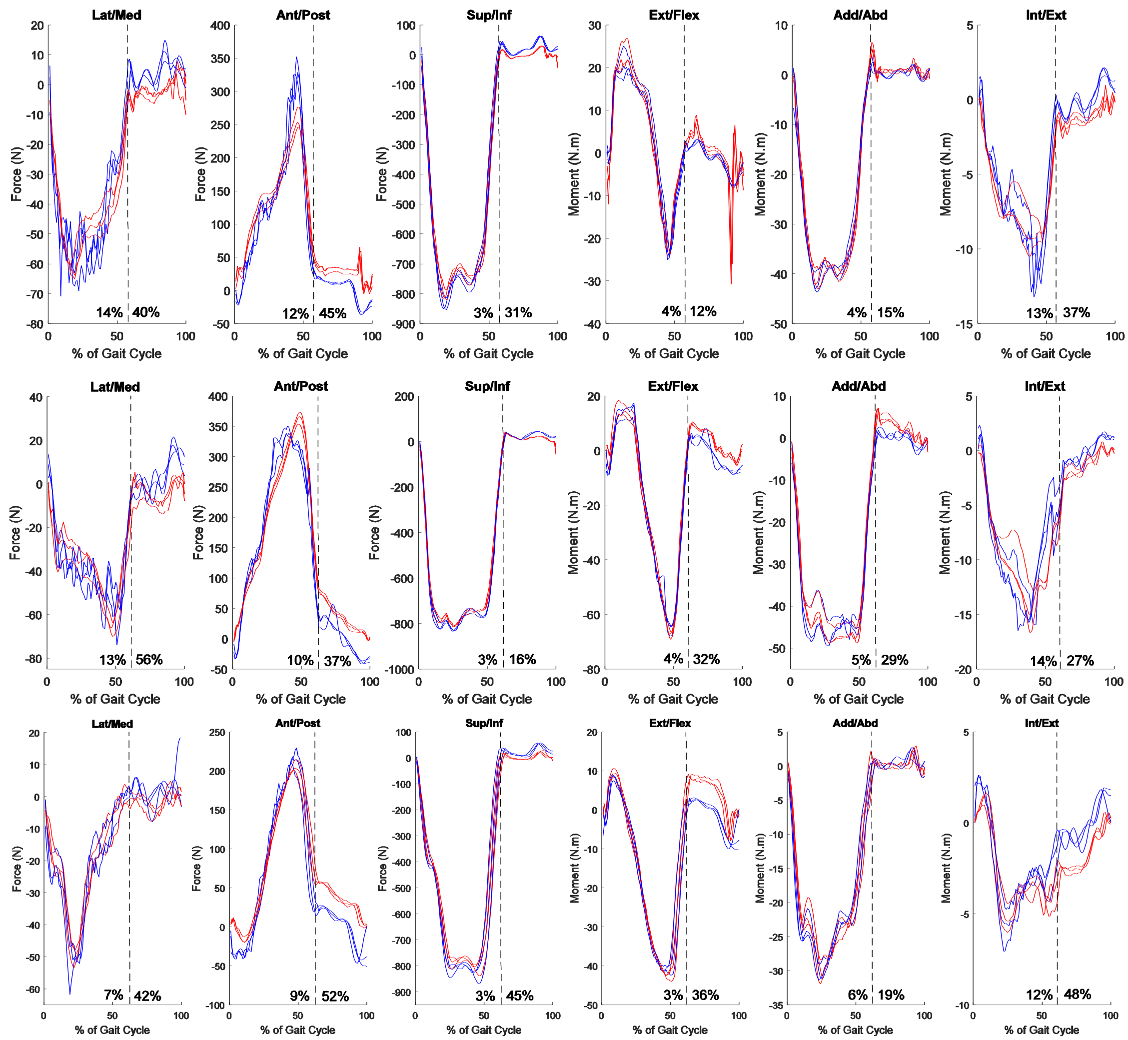

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TABLE I. Individual and overall participants demographics, amputation and prosthesis characteristics

Participant	Gender	Age (yrs.)	Height (m)	Mass (kg)	Cause of amputation	Side of amputation	Time since fitting (yrs.)	Prosthetic knee	Self-selected speed (m/s)
#1	F	56	1.63	58	Trauma	R	7	Total Knee 2000	1.18
#2	M	50	1.81	71	Trauma	L	7	Total Knee 2000	1.16
#3	M	59	1.85	83	Trauma	R	5	Total Knee 2000	0.88
#4	F	48	1.58	50	Tumour	R	1	Total Knee 2000	1.22
#5	M	41	1.77	93	Trauma	R	8	C-Leg	1.02
#6	M	26	1.78	86	Trauma	R	3	Total Knee 2000	1.15
#7	M	46	1.89	95	Trauma	L	4	C-Leg	1.00
#8	M	48	1.82	96	Tumour	R	3	Mauch Knee	0.88
#9	F	34	1.70	92	Trauma	L	2	Mauch Knee	1.51
#10	M	45	1.72	77	Other	R	4	Total Knee 2000	1.08
Mean		**45**	**1.76**	**80**			**4**		**1.11**
SD		**10**	**0.10**	**16**			**2**		**0.18**