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IEEE transactions on neural systems and rehabilitation engineering : a publication of the IEEE Engineering in Medicine and Biology Society (ISSN: 1558-0210)

Citation for the published paper:
Frossard, L. ; Hagberg, K. ; Häggström, E. et al. (2009) "Load-relief of walking AIDS on osseointegrated fixation: instrument for evidence-based practice.". IEEE transactions on neural systems and rehabilitation engineering : a publication of the IEEE Engineering in Medicine and Biology Society, vol. 17(1), pp. 9-14.

http://dx.doi.org/10.1109/TNSRE.2008.2010478

Load-Relief of Walking Aids on Osseointegrated Fixation: Instrument for Evidence-Based Practice

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Abstract—**Clinicians are currently in demand of tools enabling individual assessment during their daily practice of load-relief of walking aids. The first aim of this article is to describe a portable kinetic system that could be used to measure directly the true load applied on the residuum during assisted walking. The second aim is to present the information that can be derived from the raw loading data. The third aim is to provide an example for a participant. One active transfemoral amputee fitted with an osseointegrated fixation was asked to walk in straight level line with no aid, one stick, one and two elbow crutches on a 20 m walkway. The load-relief was measured using a six-channel transducer and recorded using a data logger. The overall loading was decreased by 2% using one stick, 5% using one crutch and by 10% using two crutches. This study presents a method that can be used by clinicians facing the challenge of prescribing and assessing walking aids to restore the locomotion of lower limb amputees in the framework of an evidence-based practice.**

Index Terms—**Gait, kinetics, osseointegration, transfemoral amputation, walking aids.**

I. Introduction

WALKING aids are commonly prescribed to lower limb amputees during and after rehabilitation [1]–[4]. For instance, transfemoral amputees fitted with an osseointegrated fixation are using two crutches, one crutch and a stick as the rehabilitation progresses until full body weight can be applied on the fixation [5]–[7]. Most of these patients are advised to use a stick during everyday locomotion although they would be able to walk independently. This clinical advice is mainly based on the assumption that walking aids can reduce normal and occasional odd loading that could potentially accelerate the fatigue of the fixation. It is also believed that a stick could reduce lateral trunk bending. Finally, it could increase stability by providing additional support. This could prevent falls and subsequent bending of the fixation [5], [6].

Manuscript received January 30, 2008; revised July 07, 2008; accepted August 28, 2008. First published December 09, 2008; current version published February 11, 2009. This work was supported in part by the Australian Research Council Discovery Project (DP0345667), in part by the Australian Research Council Linkage Grant (LP0455481), and in part by the Queensland University of Technology Strategic Link with the Industry and Institute of Health and Biomedical Innovation Advanced Diagnosis in Medical Device Grant.

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Digital Object Identifier 10.1109/TNSRE.2008.2010478

The clinical and biomechanical outcomes of assistive devices [1] have been presented by studies focusing mainly on lower limb amputees fitted with conventional socket [2], [3] and on people with various types and levels of disability [8]–[13]. However, some clinicians such as prosthetists or physiotherapists feel that relying on scientific papers is partially sufficient for an evidence-based prescription of walking aids. These clinicians are currently in demand of tools enabling individual assessment during their daily practice.

Classically, they can assess the load-relief of walking aids on the residuum using inverse dynamics equations [14], [15]. This method relies on kinematic data captured by a motion analysis system and the ground reaction forces measured by force-plates and instrumented walking aids [16]–[18]. The shortcomings of this method include, but are not limited to, the measurement of only one or two steps of walking, force-plate targeting producing altered gait, need for accurate determination of inertia of limb segments and compounded errors when involving multiple joints. Furthermore, the instrumentation of walking aids is often tedious and complex (e.g., few sensors commercially available, weight of sensors, synchronization with kinematic and dynamic systems). Finally, the collection and analysis of the data are often time-consuming and labor intensive. Consequently, this method is only partially suitable in the daily practice of most clinicians.

In principle, clinicians could also measure the load-relief using load sensors embedded into the prosthesis. Homemade transducers can be used but they could pose problems of calibration, reliability and accuracy [19]–[22]. More recently, portable kinetic systems based on low profile commercial load cell connected to recording device such as wireless modem [23] and data logger [24]–[26] have been presented. This method presents the distinct advantage to measure directly the three components of force and moment without calculations in contrast with the inverse dynamics. Furthermore, it provides more realistic information as the amputee can walk freely without being tethered by any cable. Consequently, this method appears as a relevant and practical alternative to clinicians. Unfortunately, previous studies relying on this method limited their focus on the magnitude and variability of load applied on the residuum of transfemoral amputees fitted with a socket or osseointegrated implant during walking and activities of daily living [24]–[26]. Thus, there is currently no study demonstrating the relevance of this system to determine the load-relief effect of walking aids.

Consequently, the first aim of this article is to describe an apparatus and the procedure that could be used to measure directly the true load-relief applied of the residuum during walking with no aid, one stick, one and two elbow crutches. The second aim is to present the information that can be derived from the raw loading data provided by the portable kinetic system. The third aim is to provide an example for a transfemoral amputee fitted with an osseointegrated implant.

II. Methods

A. Participant

Approximately 90 transfemoral amputees worldwide have experienced the benefits of an osseointegrated fixation [5], [27]–[29]. One of them participated in this study (Male, 40 yr, 91.5 kg, 1.78 m). The amputation was due to a trauma. The participant has been successfully fitted with an osseointegrated fixation since 1998. He could walk easily independently although he regularly uses one stick at home and outdoors as recommended by his clinicians. The participant was selected because his residuum was sufficiently short to enable the mounting of the transducer without modifying the alignment of the prosthesis. He was fully rehabilitated like most participants in previous studies focusing on walking aids. This enabled a single recording session of all the walking conditions with the same alignment of the prosthesis.

The study was approved by the Queensland University of Technology's human research ethics committee. The entire procedure was explained to the subject at which point he read and signed the informed consent form to participate in this study.

B. Apparatus

The loading was directly measured with a customized portable kinetic recording system including a transducer and a data logger. The same six-channel transducer (Model 45E15A; JR3 Inc, Woodland, CA) presented previously was used [23]–[26]. It was constructed from a solid billet of aluminium measuring 11.43 cm in diameter, 3.81 cm thick and weighing less than 800 g. Its maximum capacity was 2,273 N for long axis, 1,136 N for antero-posterior and medio-lateral axes, and 130 Nm for moments about the three axes. Accuracy was 0.1% of the maximum capacity. The output of the transducer was digitally recorded at a sampling frequency of 512 Hz and stored using an 8-bit data logger (Valitec AD128) via additional interface circuitry. It was connected to the transducer via a serial cable and attached to the subject by a waist pack. This transducer was mounted between the knee and the fixation as presented in Fig. 1. Its coordinate system was aligned so that the long axis was co-axial (compression was positive) with the fixation while the other axes corresponded to the anatomical anterior-posterior (anterior was positive) and medio-lateral (lateral was positive) directions. The rest of the prosthesis was composed of the participant's usual knee (NK-1, Nabtesco), foot (Truestep) and light running shoes. The overall weight of the prosthetic leg was 3.7 kg.

C. Procedure

First, the prosthesis including the transducer was fitted to the amputee by a prosthetist, who replicated the usual alignment. The prosthetic leg was worn approximately 15 min before recording to ensure subject confidence and comfort. Then, the subject was asked to walk at his natural pace in 20-m-long straight line using no walking aid, one stick, one and two elbow crutches. Each walking condition was repeated four times with sufficient rest in between trials to avoid fatigue.

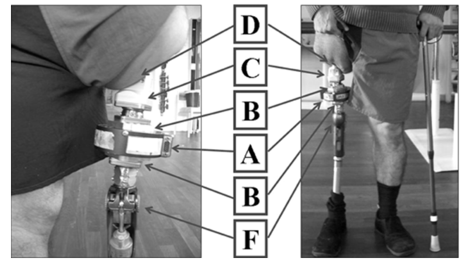

Fig. 1. Apparatus used to measure the load-relief of walking aids on the residuum of a transfemoral amputee (left: back view, right: front view). A commercial transducer (A) was mounted to specially designed plates (B) that were positioned between the adaptor (C) connected to the osseointegrated fixation (D) and the knee mechanism (F). The transducer was connected to the data logger by serial cable and attached to the subject via a waist pack.

The patient was instructed to walk as he normally would when using the different walking aids. He chose a two-point gait while walking with one stick and one crutch. He used the walking aid on the opposite side to the prosthetic leg and moved both forward together. He used a three-point gait while walking with two crutches. The walking aids and the prosthetic leg moved forward together, while the weight was taken on the sound leg. That leg was then advanced, while the weight was taken on the prosthetic leg and the crutches [30].

D. Data Analysis

The raw data generated by the transducer was preprocessed and analyzed using a customized Matlab software program (Math Works, Inc). Firstly, raw force and moment data were calibrated using a specific recording of an initial unloaded condition to remove any offset in the data and a transducer specific calibration matrix to eliminate sensor crosstalk. Secondly, the three first and the last strides recorded for each trial were discarded in order to avoid the initiation and termination of walking [31]. Thirdly, each heel contact and toe-off was detected manually using the curve of the force applied on the long axis. This allowed the calculation of the cadence expressed in strides per minute and time normalization of gait cycle (0%–100%). Fourthly, the time of occurrence expressed in percentage of the gait cycle and the magnitude of points of interest, also called local extrema, of forces and moments were determined manually for each step of the prosthetic limb. The impulse for each stride of the prosthetic limb was determined using conventional trapeze methods based on the integration of the area under the force-time curves [32]. Finally, data sets obtained for each gait cycle of the prosthetic leg were collated in one group. The speed of walking was determined based on the duration of each trial of walking recorded manually by an operator.

III. Results

A. Temporal Variables

The total number of strides considered for walking with no aid, one stick, one and two crutches was 54, 51, 50, and 49,

TABLE I
MEAN AND STANDARD DEVIATION OF THE CADENCE, THE SPEED OF WALKING AS WELL AS THE DURATION OF GAIT CYCLE, SUPPORT AND SWING PHASES DURING WALKING WITH AND WITHOUT ASSISTANCE

		No aids (Mean ± SD)	One stick (Mean ± SD)	One crutch (Mean ± SD)	Two crutches (Mean ± SD)
Cadence	(Strides/min)	43.26 ± 0.41	41.98 ± 0.4	41.95 ± 0.55	41.89 ± 0.31
Speed of walking	(m/s)	0.82 ± 0.00	0.88 ± 0.01	0.87 ± 0.00	0.89 ± 0.03
Duration gait cycle	(sec)	1.39 ± 0.03	1.43 ± 0.03	1.43 ± 0.05	1.43 ± 0.03
Duration of support	(sec)	0.87 ± 0.03	0.91 ± 0.03	0.9 ± 0.04	0.91 ± 0.04
Duration of support	(%GC)	62.83 ± 1.61	63.65 ± 1.55	62.89 ± 1.53	63.58 ± 1.88
Duration of swing	(sec)	0.52 ± 0.02	0.52 ± 0.02	0.53 ± 0.03	0.52 ± 0.03
Duration of swing	(%GC)	37.17 ± 1.61	36.35 ± 1.55	37.11 ± 1.53	36.42 ± 1.88

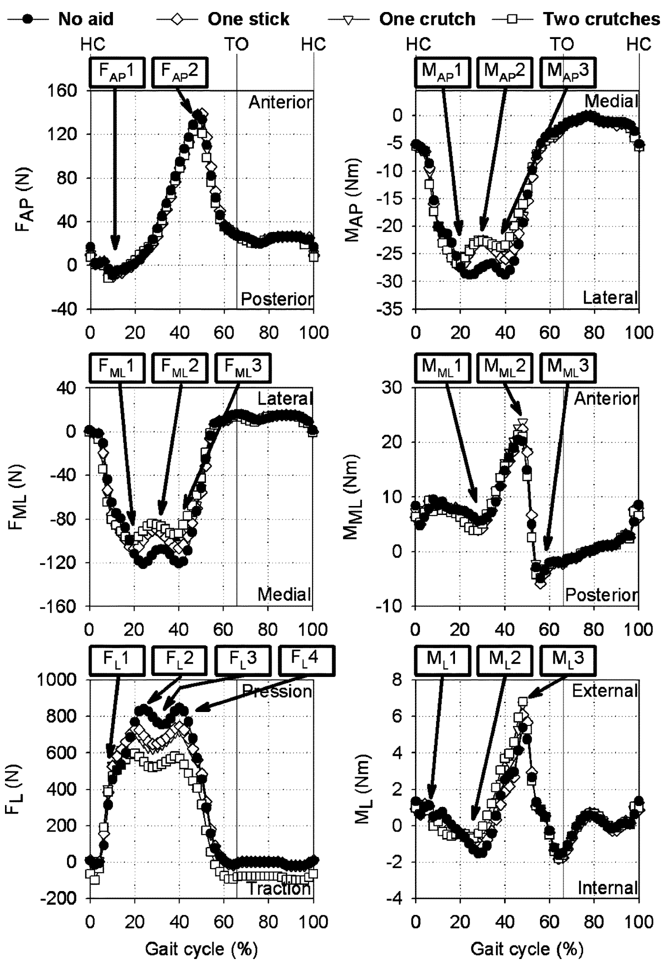

Fig. 2. Mean of forces and moments applied on the antero-posterior (FAP, MAP), medio-lateral (FML, MML) and long (FL, ML) axes during the gait cycle of walking with and without assistance. (Only one every two data points without the standard deviation is displayed for clarity). The zones pointed by the arrows identified the points of interest. HC: heel contact, TO: toe off.

respectively. The cadence, the speed of walking as well as the duration of gait cycle, support and swing phases are provided in Table I.

B. Loading Patterns

The mean of forces and moments applied on the antero-posterior (FAP, MAP), medio-lateral (FML, MML) and long (FL, ML) axes during the gait cycle are plotted in Fig. 2. Each curve represents the average of all the steps measured for the unassisted and assisted walks. The standard deviation of each curve and one of every other data points were not plotted for purposes of clarity. However, the standard deviation of each local extrema is provided in Tables II and III.

The three components of force followed a pattern that was similar to the ground reaction forces obtained with force-plates and instrumented walking aids [2], [10]–[12]. As expected, FL

TABLE II
MEAN AND STANDARD DEVIATION OF THE TIME OF OCCURRENCE EXPRESSED AS PERCENTAGE OF GAIT CYCLE (%GC) AND MAGNITUDE OF THE LOCAL EXTREMA OF THE FORCE APPLIED ON THE ANTERO-POSTERIOR (FAP), MEDIO-LATERAL (FML) AND LONG (FL) AXES OF THE OSSEOINTEGRATED FIXATION DURING WALKING WITH AND WITHOUT ASSISTANCE

	No aids (Mean±SD)	One stick (Mean±SD)	One crutch (Mean±SD)	Two crutches (Mean±SD)
	Time (%GC)			
$F_{AP}1$	14.67 ± 1.24	13.62 ± 1.60	13.58 ± 1.46	12.42 ± 1.34
$F_{AP}2$	77.71 ± 1.48	77.56 ± 1.80	77.85 ± 1.91	76.08 ± 1.72
$F_{ML}1$	37.56 ± 2.32	30.47 ± 3.11	29.30 ± 3.75	28.06 ± 3.95
$F_{ML}2$	52.76 ± 2.08	47.86 ± 2.70	47.40 ± 2.88	46.11 ± 2.89
$F_{ML}3$	65.19 ± 1.66	64.02 ± 2.35	63.79 ± 2.47	61.95 ± 2.54
F_L1	20.50 ± 1.70	18.00 ± 2.20	16.98 ± 1.77	15.79 ± 1.94
F_L2	38.17 ± 1.83	32.33 ± 1.82	31.32 ± 1.90	29.74 ± 3.12
F_L3	51.81 ± 2.08	45.30 ± 2.11	44.90 ± 2.08	44.47 ± 2.10
F_L4	64.15 ± 1.68	62.73 ± 1.71	63.38 ± 1.93	61.15 ± 2.01
	Magnitude (N)			
$F_{AP}1$	-13.35 ± 3.32	-16.57 ± 3.93	-14.49 ± 4.74	-15.57 ± 5.38
$F_{AP}2$	142.68 ± 11.77	142.51 ± 9.46	138.98 ± 9.70	132.68 ± 7.62
$F_{ML}1$	-123.34 ± 4.78	-110.68 ± 7.73	-105.09 ± 6.70	-103.77 ± 8.23
$F_{ML}2$	-104.52 ± 5.09	-87.03 ± 6.09	-83.25 ± 5.50	-81.85 ± 6.48
$F_{ML}3$	-124.79 ± 4.67	-108.95 ± 7.99	-105.38 ± 7.17	-96.87 ± 6.36
F_L1	517.84 ± 49.23	569.36 ± 42.95	552.14 ± 46.77	539.61 ± 44.64
F_L2	845.14 ± 21.40	741.99 ± 36.06	713.32 ± 33.48	690.15 ± 40.43
F_L3	748.19 ± 14.06	643.52 ± 20.99	620.87 ± 20.95	593.90 ± 26.92
F_L4	851.34 ± 17.33	750.72 ± 26.36	723.46 ± 22.47	662.14 ± 25.57

TABLE III
MEAN AND STANDARD DEVIATION OF THE TIME OF OCCURRENCE EXPRESSED AS PERCENTAGE OF GAIT CYCLE (%GC) AND MAGNITUDE OF THE LOCAL EXTREMA OF THE MOMENT APPLIED AROUND THE ANTERO-POSTERIOR (FAP), MEDIO-LATERAL (FML) AND LONG (FL) AXES OF THE OSSEOINTEGRATED FIXATION DURING WALKING WITH AND WITHOUT ASSISTANCE

	No aids (Mean±SD)	One stick (Mean±SD)	One crutch (Mean±SD)	Two crutches (Mean±SD)
	Time (%GC)			
$M_{AP}1$	36.69 ± 4.53	30.81 ± 2.80	29.84 ± 3.06	29.07 ± 2.68
$M_{AP}2$	51.71 ± 5.07	47.05 ± 3.34	47.28 ± 2.63	46.11 ± 2.93
$M_{AP}3$	63.92 ± 2.73	63.66 ± 2.99	63.99 ± 3.04	60.64 ± 3.19
$M_{ML}1$	46.50 ± 3.41	45.46 ± 2.12	43.90 ± 2.40	43.26 ± 3.26
$M_{ML}2$	75.71 ± 1.43	75.81 ± 2.07	76.27 ± 1.95	74.75 ± 1.77
$M_{ML}3$	88.58 ± 1.85	87.72 ± 2.38	88.64 ± 2.10	86.73 ± 2.11
M_L1	18.10 ± 1.41	17.40 ± 1.74	16.64 ± 2.63	15.56 ± 1.96
M_L2	45.43 ± 3.19	45.92 ± 4.01	42.29 ± 4.54	39.48 ± 4.77
M_L3	77.22 ± 1.64	76.86 ± 1.82	77.23 ± 2.04	76.03 ± 1.62
	Magnitude (N.m)			
$M_{AP}1$	-29.55 ± 1.81	-27.86 ± 2.86	-26.76 ± 2.55	-27.59 ± 3.04
$M_{AP}2$	-26.06 ± 1.81	-22.09 ± 2.45	-21.67 ± 2.44	-22.07 ± 2.34
$M_{AP}3$	-29.51 ± 1.47	-26.83 ± 2.70	-25.85 ± 2.17	-24.50 ± 1.65
$M_{ML}1$	5.00 ± 1.58	3.55 ± 1.32	3.88 ± 1.12	3.24 ± 1.35
$M_{ML}2$	21.62 ± 1.65	22.67 ± 2.06	24.35 ± 1.87	22.21 ± 1.91
$M_{ML}3$	-5.83 ± 1.15	-6.55 ± 1.28	-5.48 ± 1.22	-6.20 ± 1.13
M_L1	0.94 ± 0.35	0.66 ± 0.27	0.76 ± 0.37	0.51 ± 0.40
M_L2	-1.75 ± 0.54	-1.37 ± 0.52	-1.00 ± 0.55	-0.80 ± 0.53
M_L3	5.87 ± 0.69	6.54 ± 0.57	6.88 ± 0.63	7.07 ± 0.48

was the largest in magnitude among the three components of forces and presented two distinct peaks occurring at approximately 20% and 40% of a gait cycle, respectively, separated by a valley at approximately 30% of the gait cycle. However, it can also be observed that FL is slightly negative during the swing phase. This is due to the traction created by the gravity acting on the mass of the prosthesis when the prosthetic foot is off the floor. Consequently, the magnitude of FL varied depending on the acceleration of the prosthesis during the swing phase. Axial rotational moment (ML) was the lowest in magnitude comparing to MAP and MML because of the shorter moment arm. However, some external rotation was noticeable during the late stance phase of the gait. The fixation experienced some braking posterior forces (FAP) during early stance phase and propulsive anterior forces during the late stance phase of the gait. The participant demonstrated anterior rotational moment from early to mid stance. Posterior rotational moment was experienced when the line of action of ground reaction shifted posteriorly to the transducer during the late stance phase of the gait. The fixation consistently experienced some lateral force (FML) and rotational moment (MML) during stance phase of the gait. The magnitude of FAP, MAP, FML, and MML were somehow different from the ones that could be obtained with force-plate data. This is due to the orientation of the transducer slightly tilted laterally and posteriorly in relation to the ground (Fig. 1). This also explained why FAP and FML are not nil during the swing phase.

C. Point of Interest

A total of eighteen points of interest were studied, which represented the key features of the curve plotting force and moment data against time as presented in Fig. 2. For example, two points were identified for FAP representing the breaking phase (maximum anterior force, FAP1) and pushing phase (maximum posterior force, FAP2). The time of occurrence and magnitude of all the points in each walking condition are provided in Tables II and III.

All the points of forces and moments occurred either at the same time or earlier in the gait cycle for each assisted walking condition, except for the FAP2 and MML2 with one stick.

The maximal force applied on the antero-posterior (FAP2), medio-lateral (FML3) and long (FL4) axes corresponded to 16%, 14% and 95% of the body weight, respectively during unassisted walking. The magnitude of FAP1 and FL1 was increased by 24% and 10% using one stick, by 9% and 7% using one crutch and by 17% and 4% using two crutches. The magnitude of FAP2 was decreased for all assisted walks. The magnitude of all the other local extrema on the medio-lateral (FML1, FML2, FML3) and long (FL2, FL3, FL4) axes was reduced for each assisted walk. The average reduction of these extrema was 13% using one stick, 12% using one crutch and 20% using two crutches.

The magnitude of MML2 and ML3 was increased by 5% and 11% using one stick, by 13% and 17% using one crutch and by 3% and 20% using two crutches. The magnitude of MML3 was increased by 12% using one stick and 6% using two crutches and decrease by 6% using one crutch. The magnitude of all the other points on the antero-posterior (MAP1, MAP2, MAP3), medio-lateral (MML1) and long (ML1, ML2) axes was reduced for each assisted walk. The average reduction of these extrema was $18 \pm 10\%$ using one stick, $21 \pm 12\%$ using one crutch and $29 \pm 19\%$ using two crutches.

TABLE IV
MEAN AND STANDARD DEVIATION OF THE TIME OF OCCURRENCE EXPRESSED AS PERCENTAGE OF GAIT CYCLE (%GC) AND MAGNITUDE OF THE LOCAL EXTREMA OF THE FORCE APPLIED ON THE ANTERO-POSTERIOR (FAP), MEDIO-LATERAL (FML) AND LONG (FL) AXES OF THE OSSEOINTEGRATED FIXATION DURING WALKING WITH AND WITHOUT ASSISTANCE

	No aids (Mean ± SD)	One stick (Mean ± SD)	One crutch (Mean ± SD)	Two crutches (Mean ± SD)
I_{AP}	41.12 ± 3.47	42.32 ± 2.59	40.81 ± 3.02	40.71 ± 2.95
I_{ML}	62.37 ± 3.15	60.51 ± 3.50	58.29 ± 3.90	56.18 ± 3.58
I_L	441.50 ± 19.73	431.05 ± 16.31	418.28 ± 20.96	398.09 ± 17.36
I	447.78 ± 20.28	437.33 ± 16.88	424.29 ± 21.53	404.09 ± 17.97

D. Impulse

The mean and coefficient of variation (COV) of the norm (I) and the antero-posterior (IAP), medio-lateral (IML) and long (IL) components of impulse are provided in Table IV. As expected, IL was the largest in magnitude among the three axes. The overall loading represented by the norm of the impulse (I) was decreased by 2% using one stick, 5% using one crutch and by 10% using two crutches. However, IAP was slightly increased using one stick and decreased using one or two crutches. IML and IL axis was decreased by 3% and 2% using one stick, by 7% and 5% using one crutch and by 10% using two crutches, respectively.

IV. DISCUSSION

On one side, the proposed method has strong points that could overcome some of the shortcomings of the conventional methods using a gait laboratory. Its practical strength relies on the association of a discrete size transducer and a data logger enabling recording without cables for data transmission to fixed equipment. Thus, this intrinsic added value of this method is that it enables a direct measurement of the actual load-relief during unlimited number of steps. All combined, this method can provide a realistic description of the load-relief on the residuum of lower limb amputees fitted not only with osseointegrated fixation but also with a socket. In conclusion, this work demonstrates the relevance of the proposed method to determine the load-relief effect of walking aids. Furthermore, it appears suitable in any clinical environments (e.g., private practice, rehabilitation centre, etc.).

On the other hand, the proposed method presented one main limitation. No complementary data explaining the origin of the load-relief was collected. One can assume that the main source of relief was from the walking aids. However, other factors could modify the load profile measured by the transducer, such as the position of the trunk in relation to the residuum, modification of the step length, etc. This limitation could be alleviated by collecting simultaneous kinematic data using portable sensors such as accelerometers. This solution might be feasible provided that issues of overburdening participant with sensors can be solved.

V. Conclusions

A portable kinetic system based on a commercial transducer and a data logger allowing the measurement of the relief provided by walking aids on load applied on residuum has been presented. An example of raw results of these forces and moments as well as some of their derived information were provided for one transfemoral amputee to illustrate the capacities of this new apparatus. As the matter of fact, the overall loading was decreased by 2% using one stick, 5% using one crutch and by 10% using two crutches.

The results demonstrated that the portable kinetic recording system overcame some of the shortcomings of the conventional methods using a gait laboratory. However, additional kinematic information will be needed to further understand the loading profile.

The preliminary trial of this portable kinetic system demonstrated that it might be a suitable alternative for a wide range of clinical practices. For instance, it has the potential to participate in the decisions made by physiotherapists and prosthetists in relation to walking aids. This system could provide quantitative feedback about the actual benefit and usage of walking aids during rehabilitation program and fitting of lower limb amputees.

In conclusion, the apparatus presented here is a stepping-stone in on-board and user-friendly sensors to be used by clinicians facing the challenge of prescribing and assessing walking aids. This should help them to restore the locomotion lower-limb amputees in the framework of an evidence-based practice.

Acknowledgment

The authors would like to thank Prof. M. Pearcy (School of Engineering Systems and Institute of Health and Biomedical Innovation of the Queensland University of Technology) for his valuable contribution and feedback during the writing of this manuscript.

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Kerstin Hagberg photograph and biography not available at the time of publication.

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