Kinetics of Lower Limb Prosthesis: Automated Detection of Vertical Loading Rate

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ABSTRACT

Objective: Vertical loading rate could be associated with residuum and whole body injuries affecting individuals fitted with transtibial prostheses. The objective of this study was to outline one out of five automated methods of extraction of vertical loading rate that stacked up the best against manual detection considered as the gold standard during pseudo-prosthetic gait.

Methods: The load applied on the long axis of the leg of three males was recorded using a transducer fitted between a prosthetic foot and physiotherapy boot while walking on a treadmill for circa 30 minutes. The automated method of extraction of vertical loading rate combining the lowest absolute average and range of 95% CI difference compared to manual method was deemed the most accurate and precise.

Result: The average slope of the loading rate detected manually over 150 strides was 5.56±1.33 kN/s while the other slopes ranged from 4.43±0.98 kN/s to 6.52±1.64 kN/s depending on the automated detection method. Conclusion: An original method proposed here, relying on progressive loading gradient-based automated extraction, produced the closest results (6%) to manual selection.

Significance: This work contributed to continuous efforts made by providers of prosthetic and rehabilitation care to generate evidence informing reflective clinical decision-making.

ABBREVIATIONS

%GC: Unit if time expressed in percentage of gait cycle; %BW: Unit of vertical force (F_{LG}) expressed in percentage of body weight CI: Confidence in interval; F_{LG}: Force applied on the long axis of the leg; F_{LG1}: First loading peak during the first half of the support phase GC: Gait cycle; HC: Heel contact; Ma: Slope selected by expert a Mb: Slope selected by expert b; Mi: Method i (M1-M6) of automated detection of vertical loading rate S: Vertical loading slope; Si: Instantaneous vertical loading slope TO: Toe-off

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INTRODUCTION

Because the ability of individuals with lower limb loss to walk with a prosthesis is paramount to their quality of life, prosthetic care providers make bespoke clinical decisions intending to sustain the capacity of residuum to be fitted with a prosthesis. Allegedly, achieving satisfactory prosthetic attachment highly depends on the residuum health. This term can be defined as the holistic state of physical well-being of the residuum's distinct neuromusculoskeletal system encapsulating resected skin, nerves, muscles and bone [1-3]. The intrinsic determinants of residuum health, including mainly the length of residuum and muscles reassignment, are established during surgical amputation [4,5]. Most common extrinsic determinants of residuum health could be substantially influenced by rehabilitation specialists and suppliers of components (e.g., manufacturers, prosthetists) who facilitate control of the prosthetic joint movements and fitting of components that, altogether, ultimately pertain level of activity [6].

Importance of loading to maintain residuum health

By definition, clinical decisions around management of extrinsic determinants are more likely to have physical ripple effects on the residuum (e.g., Skin damage, muscle contracture, heterotopic bone growth, neuroma, phantom pain). These mechanical constraints constitute a loading profile broadly described as the pattern of three forces and moments applied on and around the anatomical axes of residuum over a series of gait cycles.

For example, prosthetists seek to align ankle and knee units of a lower limb so that individuals could comfortably apply half and full bodyweight while standing and ambulating, respectively. Such symmetrical loading should enable a more balanced gait that, in return, might reduce incidence of falls as well as muscloskeletal injuries of sound joints due to overuse (e.g., back pain, osteoarthritis) [3,7,8]. Choice and alignment of components for typical socket-suspended prostheses are critical to generate suitable loading profile and subsequent intra-socket pressure [9-13]. This could minimize risks of skin damages too often responsible for prosthesis abandonment. A Goldilocks loading regimen applied by bone-anchored prostheses is also critical for safe and efficient osseointegration around the implant during rehabilitation and beyond [14-16].

Under loading could lead to loosening and infection while overloading might compromise the bone–implant interface.

Limitations of inverse dynamics

Loading profile can be calculated using inverse dynamics equations providing comprehensive dynamics and kinematics information responsible for prosthetic and sound ankle, knee and hip joints kinetics [17-24]. However, this method relies on fixed equipment placed in delimited space. Ground reaction forces are collected using floor-mounted force-plates in walkways, stairs or ramps. Positions of lower limb segments within a calibrated volume are captured using 3D motion analysis systems [17,18,25,26].Valid dynamic measurements require sole contact of each foot on a force-plate that could be achieved through individualized arrangements of the starting point and/or force-plates positioning to avoid targeting and/or repetitive recording of invalid trials [17-19]. The sum of steps to be collected in a session is conditioned by the number and arrangement of force-plates and cameras [17-19]. Otherwise, shortcomings with foot placement and number of steps could be alleviated by using an instrumented treadmill provided that 3D motion capture has the capacity to record for an extended period of time. Finally, inverse dynamics calculations are potentially sensitive to accurate extraction of inertial characteristics of prosthetic components [27,28]. In addition to being resource intensive, the extraction of the loading profile using this approach is, altogether, partially reflective of the true prosthetic loading regimen applied during daily ambulation [17,18,25]. Therefore, evidence provided this way showing effects of an intervention has limited strength.

Benefits of kinetic portable devices

Alternatively, loading profile can be directly measured using state-of-the-art portable kinetic systems including compact multi-axis transducers fitted within a prosthesis (Figure 1). Such systems are capable of recording and storing or sending wirelessly the actual load applied on residuum for a virtually unlimited number of steps during unrestricted daily activities. Therefore, they should provide more ecological prosthetic loading profile with better insights into the range of critical loading characteristics [29-33].

Direct measurement of loading profile has been conducted on case-series and cohort of individuals with Transtibial

Amputation (TTA) fitted with socket-suspended prostheses [4,9-16,18,20,25,34-45]. These studies analysed force versus moment in various planes during several walking activities (e.g., walking and turning round a circle) to compare prosthetic feet, determine the effect of anteroposterior alignment perturbations on rollover and predict intra-socket pressures [34,35,46].
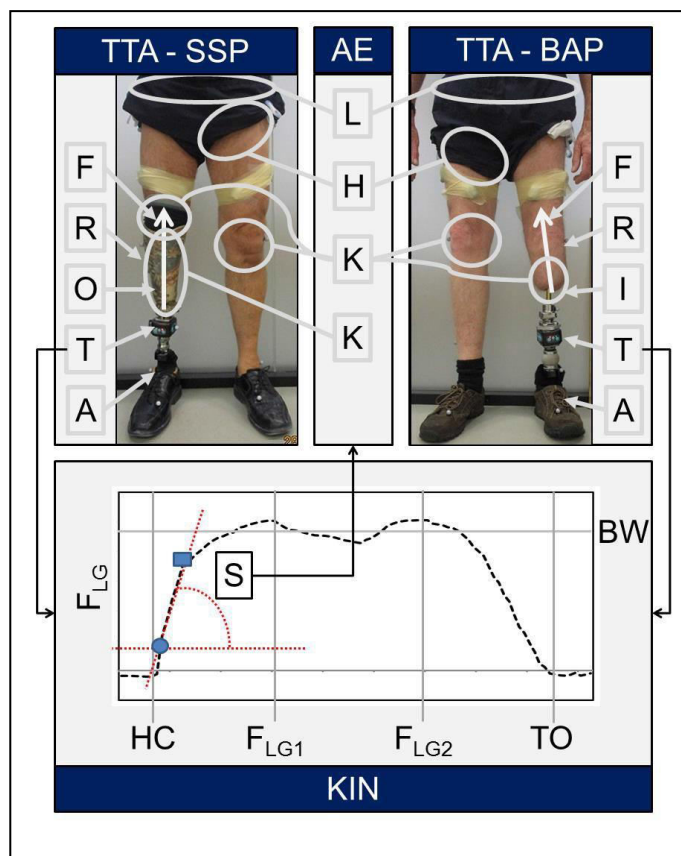

Figure 1: Overview of potential link between ankle units (A) affecting the long axis (F) of the Residuum (R) of individuals with Transtibial Amputation (TTA) fitted with either a socket (O) Suspended-Prosthesis (SSP) or a Bone-Anchored Prosthesis (BAP) through osseointegrated Implant (I), Kinetic Data (KIN) including the slope of vertical loading rate (S) of quasi-linear section of the force applied on the long axis (F_{LG}) of the residuum, measured directly by Transducer (T) that occurred between Heel Contact (HC) and first loading peak (F_{LG1}) during the first half of the support phase, and Adverse Events (AE) including skin damages of residuum and musculoskeletal injuries of Low back (L) and Hips (H) and Knees (K). TO: Toe-off, BW: body weight.

Loading profile has also been directly measured on cohorts of individuals fitted with transfemoral bone-anchored prostheses during rehabilitation program (e.g., static load bearing, use of walking aids), standardized (e.g., walking in straight line and around a circle, ascending and descending stairs and ramps) as well as unscripted daily activities (e.g., open environment, fall) [4,14-16,18,25,36-45]. These studies characterized the prosthetic loading profile using a range of variables associated with spatio- temporal characteristics (e.g., cadence, duration of Gait Cycle (GC) and support and swing phases), loading boundaries (e.g., maximum and minimum magnitude), a series of points of interest or local extremum (e.g., onset and magnitude of points of inflection between loading rate) and impulse [36-39,42,45,47,48]. Extraction of these variables for large number of steps usually generated during ecological recordings was facilitated by the semi-automated detection of gait events (e.g., heel contact (HC), toe-off (TO)) and points of interest using set loading thresholds as well as extraction of maximum or minimum loading magnitude within a time window selected manually, respectively [25,28,49-51].

Need for automated detection of loading rate

Incidentally, a single-case study differentiated the specific loading profile of transfemoral bone-anchored prostheses fitted with either mechanical or microprocessor controlled knee units using vertical loading rate [45]. Expressed in kN/s, this slope of the loading rate corresponded to a quasi-linear section of the force applied on the long axis of the residuum (F_{LG}) occurring between HC and First Loading Peak (F_{LG1}) during the first half of the support phase (Figure 1) [25]. In contrast with points of interest represented by a magnitude of loading at a given time during the support phase, the vertical loading rate provided further information by reporting the variation of magnitude between two points of calculation occurring within HC and F_{LG1} at the critical initial loading phase.

Additional information provided by vertical loading rate has been widely reported to explain lower limb musculoskeletal injuries and evidence efficiency of footwear on able-bodied [52]. Likewise, the vertical prosthetic loading rate could be used for a better understanding of the link between loading profile and residuum skin damages or development of osseointegration as well as muscloskeletal injuries of back,

sound knees and hips (Figure 1). Nonetheless, this variable is still largely overlooked when analyzing ecological prosthetic loading profile measured directly.

The underlying reasons for under reporting vertical loading rate might be found in the extraction process itself. At this stage, there is a lack of clear indications for the selection of points of calculation occurring within HC and F_{LG1} that should best represent the quasi-linear section of F_{LG}. Detection of these points must be performed somehow manually while relying on visual inspection of F_{LG} for each individual GC. Consequently, this manual selection of the slope could be subjected to high inter-raters variability. More practically, this method is simply unattainable when considering large data sets acquired during ecological assessments, reaching up to 3,000 steps in five hours recording [36-39,45].

Altogether, there is a need for elucidating vertical loading rate and slope extraction standards that will allow supplanting manual selection by specific algorithms capable of detecting automatically the slope produced during prosthetic gait. This will facilitate consistent reporting of loading rate data essential for cross-comparison of separate studies (e.g., benchmark, meta-analyses).

Objectives

The long-term aim of this work was to facilitate the development of an automated extraction of lower limb prosthetic vertical loading rate.

The main purpose of this study was to outline a worthwhile automated method to detect points of calculation of vertical loading rate produced during pseudo-transtibial prosthetic gait.

The specific objective was to compare the efficacy of five automated methods of detection against manual selection of vertical loading rate including three automated methods relying on criterion presented in the literature and two originals methods.

METHODS

Participants

The recruitment of participants was dictated by constraints of protocol to record suitable kinetic data in experimental settings at the core of study design. Between 6-10 centimeters clearance was required to provide sufficient space to fit the transducer between the transtibial socket and the ankle unit.

Recording of the large number of steps required was achieved by walking on a treadmill for up to 30 minutes. Based on discussions with clinicians, we came to the conclusion that requesting TTAs with short residuum to walk as needed could increase residuum skin damages risks (e.g., blister). Ultimately, exposure to these unnecessarily potential harms was deemed against ethical best-practice in clinical research.

Alternatively, we recruited three able-bodied males (83 ± 14.8 kg, 1.77 ± 0.1 m, 43 ± 5 yrs) using an arm-length recruitment strategy between June 2017 and March 2018. No exclusion criteria were applied for ethnicity, gender, age, weight and height or level of activity. The specific inclusion criterion included to be free of lower limb injuries or pain at the time of recording and capable to walk for 30 minutes on a treadmill using pseudo-prosthesis. Human research ethical approval was received from University Human Research Ethics Committee of the Queensland University of Technology (1600001124) and approved by U.S. Army Medical Research and Materiel Command (USAMRMC), Office of Research Protections (ORP), Human Research Protection Office (HRPO). Written consent was obtained from all the participants.

Apparatus

Participants walked with two surrogate transtibial prostheses made of a physiotherapy boot and a solid ankle cushion heel foot. The loading applied on the left leg was measured using a portable kinetic system (i.e., iPecLab, RTC, US) including a transducer fitted between the boot and foot [23]. The transducer was positioned so that the vertical axis of its coordinate system was collinear with the long axis of the leg. A spacer replaced the transducer on the right side.

F_{LG} was recorded by the transducer set at 200 Hz and sent wirelessly to a laptop nearby while participants walked at 5 km/h speed on a treadmill with a 1% incline for 25 to 30 minutes [23].

Processing

Gait events (e.g., HC, TO) were detected automatically when FLG crossed 10% of the bodyweight [29,31,32,53]. The first $10-15$ minutes of walk corresponding to circa 100 strides were discarded to avoid considering data clouded by the participants acclimation with the pseudo-prostheses. The slope of the vertical loading rate was calculated so that a high and

low magnitude indicated a steep and flat slope for six methods (M1-M6), respectively.

Table 1: Criterion for automated detection of the slope of vertical loading rate (S) of during the first half of the support phase for each of the six methods compared including three automated methods relying on criterion presented in the literature (M2, M3, M4) and two originals methods (M5, M6). Y: Yes, N: No.

An overview of the criterion for automated detection of the slope of vertical loading rate including the section of the loading rate used to calculate the slope, the requirement for normalization of load with bodyweight (Newtons vs %BW) and time of GC (Seconds vs %GC) as well as supporting reference for each of the six methods compared is provided in (Table 1). In M1, the slopes was the average of the slopes selected by two experts (Ma, Mb) who manually identified a section of F_{LG} expressed in Newtons over time expressed in seconds they deemed the most relevant, as described in Frossard et al., while characterizing loading rate of transfemoral bone-anchored prostheses fitted with different components during walking [45]. In M2, the slopes were detected automatically for the section between 20% and 80% of the maximum F_{LG} expressed in %BW over time expressed in %GC, as described by Williams et al., while characterizing loading rate on able-bodied during running [51]. In M3, the slopes were detected automatically for the section within the first 20 ms of F_{LG} expressed in Newtons over time expressed in seconds, as described by Revill et al., while characterizing loading rate on able-bodied during walking [50]. In M4, the slopes were detected automatically for the section between 200N and 90% of the maximum F_{LG} expressed in Newtons over time expressed in seconds, as described by Lieberman et al., while characterizing loading rate on able-bodied during running [49]. In M5, the slopes were calculated between HC and F_{LG1} expressed in %BW over time expressed in %GC, as described by Frossard et al., while characterizing loading profile with transfemoral bone-anchored prosthesis during walking with aids and fall [39,44]. Purposely designed for this study, M6 relied on progressive loading gradient-based automated extraction method. The slopes were calculated between two points on FLG occurring when the Instantaneous Slopes (Si) were superior and inferior to 15% of the maximum gradient between HC and F_{LG1}, respectively. Si was calculated as the numerical first derivative of the F_{LG} expressed in %BW with respect to time expressed in %GC.

Analysis

The efficacy of a Given Method (Mi) was reported using the average and one standard deviation of all slopes detected for the three participants combined. M1 was considered as the gold standard. Automated methods were compared against M1 using the average and 95% Confidence Interval (CI) of the difference and represented in a Bland-Altman plot. A positive or negative difference indicated that the slope detected automatically was flatter or steeper than M1, respectively. The automated method combining the lower absolute average and range of 95% CI difference was deemed the most accurate and precise.

RESULT

A total of 150 gait cycles including approximately 50 cycles per participant were considered for analysis. The cadence, duration of GC and support phase was 46 strides/min, 1.32±0.11 s, 0.80±0.08 s or 61±6 %GC, respectively. An

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overview of the mean pattern of F_{LG} during the whole support phase is provided in (Figure 2).
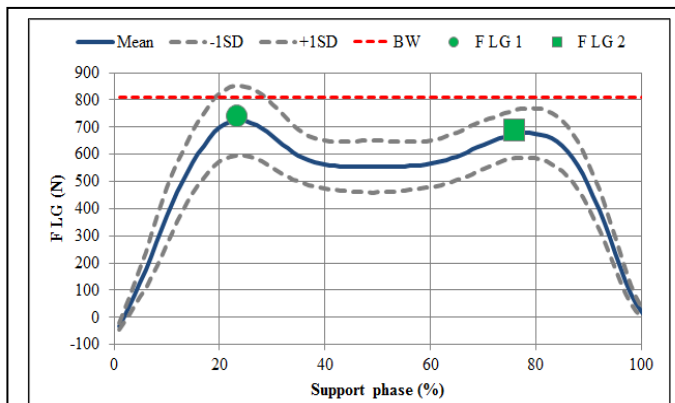

Figure 2: Mean and standard deviation of force applied on the long axis of the leg (F_{LG}) as well as mean onset and magnitude of F_{LG1} and F_{LG2} measured by transducer for 150 gaits cycles. BW: average body weight.

Figure 3: Mean and standard deviation of slopes of vertical loading rates generated with each of the six methods of detection (M1-M6) (a) as well as 150 slopes, mean and 95% confidence intervals of difference between M1 considered as gold standard and M2 (b), M3 (c), M4 (d), M5 (e) and M6 (f) represented by solid and dashed lines, respectively.

The average slope was 5.56±1.33 KN/s for M1, 6.52±1.64 KN/s for M2, 5.13±1.94 KN/s for M3, 6.11±1.65 KN/s for M4, 4.43±0.98 KN/s for M5 and 5.24±1.41 KN/s for M6.

The average slopes detected with M2 and M4 were 15% and 9% steeper while the ones with M3, M5 and M6 were 9%, 26% and 6% flatter than M1, respectively.

Mean difference with M1 and 95% CI was -0.95 [-1.92, 0.02] kN/s for M2, 0.44 [-2.37, 3.25] kN/s for M3, -0.54 [-2.02, 0.94] kN/s for M4, 1.13 [-0.68, 2.95] kN/s for M5 and 0.32 [-0.79, 1.43] kN/s for M6 (Figure 3b-f).

The ranking of the methods by decreasing absolute average and range of 95% CI differences combined was: M6, M2, M4, M3 and M5.

DISCUSSION

Key results

The key outcomes of the study were:

• The average slopes of vertical loading rate ranged between 4.43±0.98 kN/s to 6.52±1.64 kN/s for M2 to M5, respectively.

• The difference between the manual (M1) and the automated (M2-M6) detection of slopes of vertical loading rate ranged between -0.96 kN/s and 1.13 kN/s for M2 to M5, respectively.

Limitations

A clear limitation of this work was the use of an instrumented physiotherapy boot by able-bodied to produce a pseudo-prosthetic gait rather than actual loads applied by individuals with TTA during prosthetic gait.

Another limitation was the appraisal of the differences in magnitude of slopes between manual and automated methods of detection. Altogether, the differences with M1 seemed generally low for each method based on basic algebraic interpretations. Regardless of the possible statistical significance of the differences, a more comprehensive understanding was limited since there is little evidence showing how minimal clinically important differences in vertical loading rate translate into noticeable outcomes for patients (e.g., comfort score, skin damage) as highlighted previously [54,55].

Interpretation

Clearly, relying on easily detectable points of inflection, such as HC and F_{LG1}, generated the worse results with 26% difference between M5 and M1. This confirmed that vertical
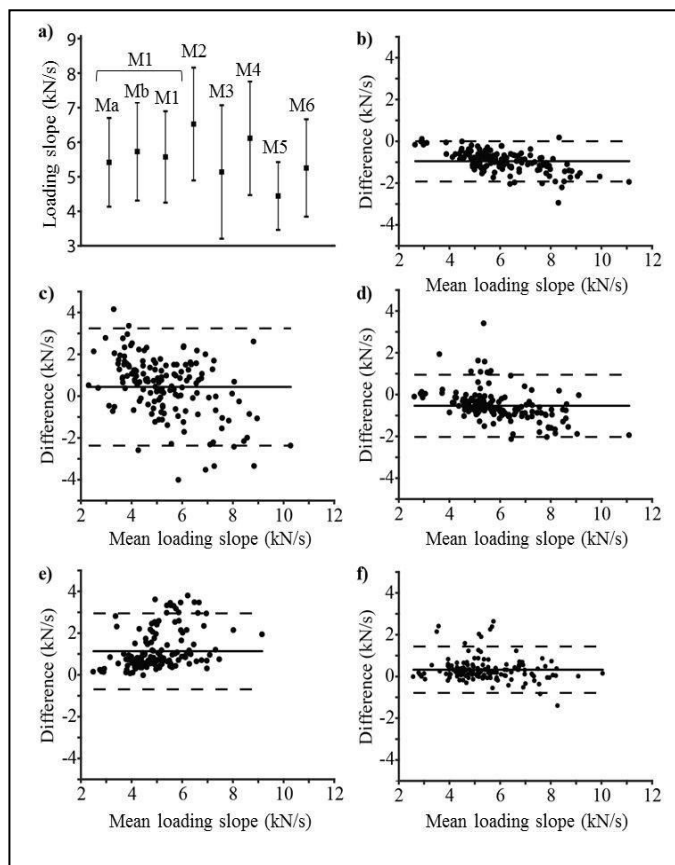
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loading in pseudo-prosthetic gait generates transient intermediate loading phases before and after the loading rate instead of a linear progression between HC and F_{LG1}.

Methods relying on previously established selection criterion generated acceptable differences of less than 15% for M2-M5. This indicates that set vertical loading thresholds extracted from dynamics studies on able-bodied during walking or running might only be partially transferable to prosthetic gait.

Furthermore, this study revealed that M6 produced the smallest difference (6%) compared to M1, indicating that a progressive loading gradient-based automated detection of vertical loading rate during pseudo-prosthetic gait might be worthwhile. Finally, comparison with existing literature showed that, interestingly, F_{LG} applied by the participants presented similar features as F_{LG} applied by actual individuals fitted with transfemoral prostheses [38,39,45]. For instance, the force applied during the push off phase represented by F_{LG2} (e.g., 76±6%SUP, 689.73±85.31 N or 86±8 %BW) was 52.45±79.12 N or 7±10 %BW smaller that F_{LG1} (e.g., 23±3%SUP, 742.18±133.90 N or 92±15 %BW). However, further comparison with loading rates published in the literature was challenging due to the discrepancy in measurements and reporting of the data. Frossard et al., showed that the loading rate on the long axis of a transfemoral osseointegrated implant during the initial part of a fall was 0.34 N/ms [44]. Frossard et al., indicated that the slope of the loading rate applied on long axis of osseointegrated implants by transfemoral bone-anchored prostheses was 70.56±1.86 deg when walking [45]. Revill et al., reported that the peak vertical force loading rate applied by abled-bodied during barefoot walking ranged between 100 and 120 BW/s.

Generalisability

Further generalization of the results was also impeded by the typical intrinsic limitations of this small case-series study. A reasonably large range of body mass and height within the group of participants should affect stride length and induced some variability in vertical loading patterns [37,38]. Nonetheless, other confounders that could potentially add variability were controlled as only able-bodied males walked with surrogate prosthesis at steady pace using a treadmill.

Future studies

The next logical step of this work will be to extract vertical loading rate applying M6 for larger cohorts of individuals with TTA performing actual daily activities [37,38,53,56]. Further longitudinal studies could improve the robustness of the algorithm proposed in M6 by considering cohorts with various lengths of residuum, body weights and functional classification. Further evidence of clinical utility of M6 could be achieved by additional cross-sectional studies establishing how vertical loading rate recorded with prosthetic components various anthropomorphic designs, alignments and methods of attachment (e.g., socket suspension, bone-anchorage) could be associated with particular adverse events affecting residuum and overall health (e.g., skin damages, development of osseointegration, injuries of sound joints) as well as participants' experience (e.g., comfort, satisfaction) [1,43,44,47,48,57].

CONCLUSION

This study provided critical technical information to report the efficacy of several methods of extraction of vertical loading rate for the first time. Indeed, an attempt to outline a worthwhile automated method relying on progressive loading gradient-based detection of points of calculation was shared.

This study should be considered as a stepping-stone in the broad developments of automated characterization of prosthetic loading profiles of individuals with lower limb amputation. More particularly, this study could facilitate analysis of large ecological kinetic datasets obtained during rehabilitation and beyond. We will argue that extraction of loading rate as suggested could be used by prosthetic care providers to guide prosthetic adjustments aiming at maintaining residuum health including choices, fitting and alignment of components. In return, the knowledge of loading rate following these interventions could assist rehabilitation specialists to make differential diagnosis of most common issues compromising the residuum health such as skin damages, muscle contracture or heterotopic bone growth.

Altogether, this work contributed to the on-going efforts made by all providers of rehabilitation and prosthetic care to generate ever better evidence informing reflective clinical decision-making.

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ETHICAL ADHERENCE

Human research ethical approval was received from University Human Research Ethics Committee of the Queensland University of Technology (1600001124) and approved by U.S. Army Medical Research and Materiel Command (USAMRMC), Office of Research Protections (ORP), Human Research Protection Office (HRPO).

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