**Quasi-stiffness of the knee joint is influenced by walking on a destabilising terrain.**

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***Abstract***

***Background***

Predictive models have been devised to estimate the necessary quasi-stiffness that a transfemoral prosthesis should be set to aligning the body and gait parameters of the user. Current recommendations exist only for walking over level ground. This study aimed to ascertain whether walking across destabilising terrain influences the quasi-stiffness of the knee joint thus influencing prosthetic engineering.

***Methods***

Ten healthy males (age: 25.1 ± 2.5 years; mean ± sd, height: 1.78 ± 0.05m, weight: 84.40 ± 11.02kg) performed 14 gait trials. Seven trials were conducted over even ground and seven over 20 mm ballast. Three-dimensional motion capture and ground reaction force were collected. Paired samples t-tests and Wilcoxon signed ranked test compared variables including; quasi-stiffness, gait speed, stride length and stride width.

***Results***

Quasi-stiffness (*d* = 0.562, *P* = 0.001) and stride width (*d* = 0.909, *P* < 0.001) were significantly greater in the destabilising terrain condition. Gait speed (*r* = -0.731, *P* = 0.001) was significantly greater in the control condition. No significant difference was seen in stride length (*d* = 0.583, *P* = 0.016).

***Conclusions***

An increase in quasi-stiffness when walking across destabilising terrain was attributed to a magnified shock absorption mechanism, facilitating an increased flexion angle during the stance phase. This causes a lower centre of mass resulting in the musculoskeletal system having to produce a greater knee extensor moment to prevent the knee collapsing. Therefore, transfemoral prostheses should be tuned to apply increased extension moments if ambulation is to occur on a destabilising terrain.

**Key words:** Quasi-stiffness; gait analysis; knee; prosthesis; transfemoral

***1.0 Introduction***

In 2005, 1.6 million individuals were living with the loss of a limb in the United States with this number predicted to increase to 3.6 million by the year 2050 [1]. It has also been reported that 20% of individuals aged over 70 years old require some sort of amputation [2,3]. With particular regard to the lower limb, approximately 5000 amputations are performed in England each year [4] with those being classified as transtibial or transfemoral in-line with whether the amputation was performed below or above the knee, respectively [5,6]. In comparison to transfemoral amputees, transtibial amputees cope better with tasks of daily life [5] which can be attributed to an intact knee joint [7]. This is due to our current inability to replicate the complex functionality of a biological knee joint [8]. Specifically, more proximal amputations have larger consequences on an individual’s ability to ambulate and therefore their quality of life [6,9].

Recent developments of transfemoral prostheses (TFP) are more capable of accurately mimicking the functions of the human knee joint when ambulating across even terrain[[1]](#endnote-1). They are, however, unable to adapt the production of force to meet the demands of walking in different situations. Consequently, a force deficit is often encountered restricting the user’s ability to ambulate efficiently. In fact, TFP users walk around 40% slower than individuals with intact limbs [9]. As a result of this, the majority of transfemoral amputees prefer to accept a life of decreased mobility by opting to use a wheelchair [10]. Recently, the concept of quasi-stiffness or “dynamic stiffness” has been used to characterise the spring-like behaviour of lower-limb joints during walking [16]. Quasi-stiffness can be defined as the stiffness of a spring that best mimics the overall behaviour of a joint during a locomotion task. One should note that the quasi-stiffness of a joint explains how a joint functions during a locomotion task or phase, distinguishing it from the passive and active stiffness of a joint defined as a specific function of angle and time [16]. The concept of quasi-stiffness applies particularly well to the knee joint during stance phase of walking, where a substantial moment is applied to compliantly support the body weight.

Ambulating over uneven terrains is a task encountered in daily life [11]. Changes seen in surface characteristics, such as; compliance, obstacles, incline or decline must be mediated during walking which places stress on the musculoskeletal system and therefore demands that constant adaptations are made [12-14]. Grasping these adaptations due to surface perturbations can highlight the functional demands of bipedalism in a range of scientific disciplines [15]. Moreover, gait adaptations become difficult for amputees who utilise TFP due to artificial knee joints being less competent in comparison to human knees [16]. Therefore, developments in design and tuning of TFPs should be based on each individual user [16], with the ultimate macroscopic goal being full functionality on a multitude of terrains. An intricate understanding of healthy human knee joint biomechanics when walking across different terrains is needed to make accurate emulation a possibility.

A common finding in the literature [17-19] asserts that the spring-like behaviour of lower-limb joints in the sagittal plane plays a vital role in governing human locomotion [16], therefore further research in this area could hold the answer to implementing greater adaptability. It is therefore postulated that this behaviour must be implemented into the design and tuning of artificial knee joints to allow amputees to experience a more natural gait pattern [16,20,21]. Moreover, adopting a pragmatic approach to determine rudimentary gait models is seen to be an accepted method of accurately distinguishing the mechanical functions of the lower limb during walking [17-19].

Quasi-stiffness (QS) is thought to quantify the spring-like behaviour of the knee joint [22][[2]](#endnote-2). QS represents the coordination between a joint moment and angular displacement, and thus serves as a quantity that portrays a neuromechanical relationship [23]. It is, therefore, recommended to investigate QS when studying the knee during gait [24-26] due to a large extension moment in the sagittal plane being applied in order to avoid the knee from collapsing in the weight acceptance phase [16,27]. The likelihood of the knee joint collapsing during this phase is heightened due to the knee joint exhibiting a shock absorption mechanism [28,29] which momentarily results in a lower body centre of mass by virtue of increased knee flexion (see figure 1). Consequently, the distance of the sagittal knee moment arm is increased and therefore, the musculoskeletal system is obligated to produce more force to maintain support of body weight [30].

*Figure 1.* Diagram of shock absorption mechanism during the initial contact phase of gait (diagram taken from Moyer, Ratneswaran, Beier & Birmingham [31]).

A fundamental problem is that no guidelines exist to which artificial knee joint stiffness should be set. Prosthesis designers generally use mean kinematic and kinetic data of healthy subjects [17,32,33] which is dependent on the sample population used. This approach also leads to large discrepancies in the default stiffness used by manufacturers which have been observed to range from 50 Newton-metres per radian (Nm·rad­-1) to 430 Nm·rad­-1 [32,33] resulting in great differences between the set stiffness and the required stiffness of the user[[3]](#endnote-3).

Shamaei et al. [16] attempted to remedy this problem by devising mathematical models to estimate the required knee joint QS over level ground according to the user’s body and gait parameters. Observed minimum and maximum QS values were reported to be 98 Nm·rad­-1 and 565 Nm·rad­-1 along with a mean QS of 284 Nm·rad­-1, illustrating the importance of tuning a TFP according to the user. Participants exhibited equal QS for the flexion and extension phases (spring-like behaviour) of the stance phase of gait in the sagittal plane between the non-dimensional gait speeds of 0.174 Fr and 0.247 Fr with a mean speed of 0.213 Fr. Therefore, the use of a single torsional spring equal to the QS of the weight acceptance phase was proposed as a suitable method of tuning a prosthesis over level ground.

The purpose of this study was to ascertain whether mathematical models that predict knee joint QS based on body and gait parameters of speed, stride length, and stride width over level ground require adapting when walking across a destabilising terrain. Findings could be implemented into TFP design and tuning.

***2.0 Method***

***2.1 Subjects***

Participants were provided with an information sheet in addition to verbal explanation of the testing procedure. Informed consent was then obtained. A health history questionnaire was also completed. The current study was approved by the local University Ethics Committee.

Ten healthy males (age: mean 25.1 ± 2.5 years ±; height: mean 1.78 ± 0.05 metres (m); weight; mean 84.4 ± 11.0 kilograms) participated in the current study. Participants were not considered for testing if they had a body mass index of >30 kg·m-2 or had sustained any musculoskeletal injuries in the last five years that had influenced their walking gait[[4]](#endnote-4),[[5]](#endnote-5).

***2.2 Experimental Protocol***

2.2.1 Equipment Set-up

Kinematics were collected using a ten camera motion capture system (MX- T-series; Vicon, Oxford, UK).Cameras sampled at 250 Hz, greater than that of previous literature investigating walking tasks [34,35]. A force plate (FP; 9281B, Kistler Instruments, Hampshire, UK) sampling 1000 Hz [36].

2.2.2 Testing Procedure

Participant’s body height and mass (Seca, Hamburg, Germany) were recorded. Leg length was determined in-line with McClymont, Pataky, Crompton, Savage, and Bates [37] enabling the calculation of the participant’s Froude number (Fr) for each trial [38] whereby gait speed is normalised to participants’ leg length [16,38][[6]](#endnote-6).

Participants wore tight lycra shorts and vest along with standardised footwear (Nomis, Australia). Twenty-two retro-reflective calibration markers were placed at anatomical landmarks along with four tracking clusters. The lower-limb model consisted of seven segments; the pelvis and both thighs, shanks and feet. Tracking markers for the pelvis were made up of anterior and posterior superior iliac spines (ASIS; PSIS) and iliac crests [39]. The thigh and shank segments were tracked by semi-rigid clusters consisting of three markers, with static markers placed on the medial and lateral femoral condyles, and medial and lateral malleolus. The foot segment was tracked by markers located at the calcaneus and the head of the second and fifth metatarsals.

*Figure 2.* Marker placements

Tracking markers constituted the technical coordinate system for each segment [40]. A static calibration was completed for each participant prior to any trials being performed. This CAST based modelling approach was used due to its suitability for providing estimates of internal joint forces and kinematic parameters [32,33,39].

After a self-selected warm-up of approximately 5 minutes, five familiarisation trials were completed to facilitate the adaptation of their gait pattern for ambulating across destabilising terrain. A total of 14 acceptable gait trials were then recorded per participant, seven trials walking on a carpeted concrete floor and seven over destabilising terrain [34]. Gait trials were performed at the participant’s preferred gait speed as individuals prefer to walk at a speed determined by their body size [41,42]. Trials were rejected if the FP was targeted, an unnatural gait pattern was apparent or if a trip occurred. The order at which participants performed each condition was randomised.

For the condition trials, a 2.4 x 0.6 metres (m) walkway was filled with ballast (coarse gravel of varying diameter up to 4 cm), in-line with the parameters set by Böhm and Hösl [34]. A specific section of the walkway was bolted to the FP and housed stones separately to the rest of the walkway, preventing stones from passing on and off of the FP.

*Figure 3.* Destabilising terrain walkway

Dispersion through the ballast to the FP has been demonstrated to have no significant effect on force data conceding that the centre of the plate is loaded, be it in normal or shear directions [34,43]. Therefore, a trial was discarded if the centre of pressure (CoP) was within 10 centimetres (cm) of the edge of the FP [34][[7]](#endnote-7) [[8]](#endnote-8).

***2.3 Data Analysis***

2.3.1 Video Analysis

Vicon Nexus (Vicon, Oxford, UK) was used to recorded, label and fill gaps in the marker trajectories. A Cartesian coordinate system was used whereby represented the medial/lateral position, represented the anterior/posterior position and represented the vertical position of markers, relative to the origin of the global coordinate system.

A model in Visual 3D (C-Motion, Germantown, USA) was used to adjust the height of the force structures, and therefore the CoP), for destabilising terrain trials. This allowed the height of the CoP to be raised to the point in which the foot actually made contact with the destabilising terrain as opposed to level of the FP itself. The vertical offset of the force structure was set to reside 4cm lower than the calcaneus marker at heel strike. Inverse dynamics was utilised to obtain net joint moment data [44]. Gait speed was determined in-line with Jor’dan et al. [45]. Fr was calculated using gait speed, gravitational constant and leg length (equation 1; [16]). Knee joint angle was determined by assessing the angle of the shank segment relative to the thigh segment in the sagittal plane. Stride length was acquired by calculating the displacement from right-heel strike to right-heel strike (one complete gait cycle). Stride width was obtained in-line with Salazar-González et al. [46].

|  |  |  |
| --- | --- | --- |
|  | Fr = | (1) |

Raw marker trajectories were filtered at 7Hz and force data at 20Hz using a Butterworth low-pass filter [47,48]. Sagittal knee joint moments and angles, gait speed, stride length and stride width were then calculated before being exported.

Joint moment and angle data were extracted from the weight acceptance portion of the stance phase of the gait cycle. This stage of the gait cycle includes initial heel contact, primary loading and the mid-stance (see figure 2). QS for the flexion (QSf) and extension (QS­e) phases of the weight acceptance phase was calculated in-line with Shamaei et al. [16] (see figure 3)[[9]](#endnote-9),[[10]](#endnote-10). This method was chosen over fitting a single line of best fit to the weight acceptance phase as the extension phase of the gait cycle occurs over a longer time period which may skew results [16]. Coefficient of determination (*R*2) values were calculated[[11]](#endnote-11).

*Figure 4.* Schematic of gait cycle phases (diagram taken from Shamaei et al. [16]).

*Figure 5*. Example of the moment (N.m; Y-Axis) angle (rad; X-Axis) curve when ambulating across level ground with lines of best fit (letters represent gait events stated in Figure 4.). Quasi-stiffness is calculated from the best fit line for the moment-angle curve for a-b for the flexion stage (QSf), and b-d for the extension phase (QSe) of weight acceptance.

2.3.2 Statistical Analysis

Data were analysed using SPSS v.22 (IBM, New York, USA). Skewness/kurtosis ratios of each variable was analysed visually from histograms and Q-Q plots. Fr data were found to be the only variable not normally distributed. Extreme outliers were removed and replaced by using the estimation-maximisation technique [49]. Descriptive statistics were calculated before paired samples t-tests were conducted on normally distributed variables while a Wilcoxon signed ranked test was performed on the non-parametric variable.Due to several tests being conducted on one data set, a Bonferroni adjustment was performed to lower the likelihood of a Type I error [50]; adjusted *a*-value was reduced from 0.05 to 0.013. To obtain effect sizes, Cohen’s *d* was calculated for paired samples t-tests and Pearsons *r* was calculated for Wilcoxon signed ranked test [51]. For Cohen’s *d,* an effect size of <0.20 was interpreted as trivial, 0.20 to 0.49 was interpreted as a small effect, 0.50 to 0.79 was interpreted as a medium effect and >0.80 was interpreted as a large effect [52]. For Pearson’s *r*, an effect size of 0.1-0.29 was considered a small effect, 0.3-0.49 was considered medium effect and >0.5 was considered a large effect [51].

***3.0 Results***

There was a significant increase in QS of 12.49 Nm·rad­-1 between the even and destabilising terrain conditions (*t*(29) = -3.638, *d* = 0.676, *P* = 0.001). There was a significant decrease (*z* = -3.269, *r* = -0.597, *P* = 0.001) of 0.05 Fr in gait speed between the even and destabilising terrain conditions. There was no significant difference in stride length (*t*(29) = 2.570, *d* = 0.477, *P* = 0.016), whilst there was a significant increase (*t*(29) = -7.188, *d* = 1.335, *P* < 0.001) 2.44cm in stride width between the even and destabilising terrain conditions (see table 1).

*Table 1.* Mean (SD) QS (Nm·rad­-1), Walking speed (Fr), Stride length (m) and Stride width (cm) data.

|  |  |  |
| --- | --- | --- |
|  | Destabilising terrain | Even terrain |
| QS | 69.88 (22.68) | 57.39 (21.74) |
| Walking speed | 0.16 (0.09) | 0.21 (0.10) |
| Stride length | 1.30 (0.30) | 1.50 (0.15) |
| Stride width | 14.27 (3.31) | 11.83 (1.86) |

The graphical data clearly shows the even terrain required a very similar level of quasi-stiffness in both the flexion and extension portions of the stance phase. Once participants were walking on a destabilising terrain it was noticeable the extra quasi-stiffness required during the weight-absorption, or knee flexion phase initially following foot contact.

*Figure 6.* Graphed mean (SD) data of QSf and QSe between even and destabilising terrain.

The location of the CoP during destabilising terrain trials might have been less accurate due to the ballast moving relative to the FP. Previous work had supplied evidence indicating that CoP location is insignificantly affected by ballast being placed on the FP providing certain criteria were met [43]. The offset of the vertical CoP position, however, must be manually inputted which is difficult as the foot sinks in to the stones during the weight acceptance phase (Gates et al., 2012). This would influence moment data by altering the distance of the sagittal knee moment arm, although the magnitude of effect is difficult to predict. Therefore, a post-hoc analysis was conducted in an attempt to ascertain how much moment data changes when the vertical position of the CoP is set to different heights. The analysis was conducted on the data of one participant chosen at random. Data was extracted from the same frame across the different offsets. Specifically, moment data was obtained with the CoP height set to; 0cm (floor level), 6cm, 7cm and 8cm. A difference of 0.26% in sagittal knee joint moment data was calculated between setting the CoP at floor level and 8cm, whilst no difference was observed between 6cm, 7cm and 8cm (see Table 2). Due to the combination of; large significant differences in mean QS between conditions (Table 1) and the small differences in moment data between vertical CoP offsets, it can be postulated that any minor error in CoP offset would result in the same finding. Devising a standardised method, however, of setting the vertical CoP location when walking on destabilising terrain is still warranted.

*Table 2.* Sagittal knee moment (Nm) data calculated with different force structure heights.

|  |  |  |  |
| --- | --- | --- | --- |
| 0cm | 6cm | 7cm | 8cm |
| 75.12 | 74.93 | 74.93 | 74.93 |

***4.0 Discussion***

The purpose of this study was to ascertain whether mathematical models that predict knee joint QS based on body and gait parameters over level ground require adapting when walking across a destabilising terrain. Results showed greater QS when walking on destabilising terrains, with walking speed decreased and stride width increased, while no significant change in stride length was observed.

QS was found to be significantly higher when walking on the destabilising terrain compared to even ground. This is in-line with the findings of Voloshina and Ferris [53] who asserted that subjects exhibited significantly greater leg stiffness when running over destabilising terrain in comparison to level ground. It should be noted, however, that Voloshina and Ferris [53] used GRFs instead of joint moments [16] to plot against knee joint excursion in order to calculate leg stiffness. This approach disregards the changes in distance of the sagittal knee moment arm that could be brought about by perturbations of the destabilising terrain. Furthermore, it has been postulated that the primary cause for this change in leg stiffness is due to alterations in lower limb posture. Specifically, where the foot contacts and parts with the ground with greater flexion at the knee lowering the body centre of mass [53] and increasing stability, as would an increase in stride width. This would reinforce conclusions drawn by Blum et al. [54] showing that an adaptation mechanism exists whereby the knee is closer to the middle of its range of motion as opposed to one end of it where fewer adaptations can be made. In addition, the finding of greater QS combined with previously observed lower body centre of mass when ambulating across uneven terrain [53,54] is evidence to suggest that the shock absorption mechanism [28,29] is magnified when ambulating across an uneven surface. Therefore the large QS can be attributed to an increased flexion moment being applied resulting in greater peak knee flexion. Consequently, a large extension moment is required during the extension stage to overcome the increased sagittal plane moment arm and stop the knee from collapsing [16,27]. These data implied that TFP should be individually tuned to apply an increased extension moment if ambulation is to occur on a destabilising terrain.

These findings confirmed that QS predictive stature-based models produced by Shamaei et al. [16] are only suitable for firm ground walking. Models based upon an individual’s body parameters alone must be adapted when walking across a destabilising terrain. Furthermore, the work of Shamaei et al. [16] offered the best model to predict knee joint QS with reported average error values of 9% for stature-based models and 11% for models based on both stature and gait parameters. Therefore, the estimated QS is only recommended for implementing into individual’s TFP design when walking over an even surface.

Results for QSf and QSe during the control condition were found to be nearly identical when ambulating over even terrain, which can be defined as spring-like behaviour (see Figure 4). Such findings reemphasized the work of Shamaei et al. [16] and Shamaei and Dollar [55] who reported a similar finding when participants walked at their preferred gait speed. Specifically, results came about with a mean Fr of 0.210. Shamaei et al. [16]stated a mean preferred non-dimensional gait speed of 0.213Fr, only 0.003Fr greater than that of the current study. Proposals have been made on the strength of this spring-like behaviour claiming that a single torsional spring with stiffness equal to the QS of the weight acceptance phase may be an appropriate method of tuning the stiffness of artificial knees [16]. When walking over the destabilising terrain, however, QS was much greater during the flexion stage compared to the extension stage (see Figure 4). Such a large discrepancy makes the use of a single torsional spring to tune prosthesis QS a less viable option when walking over a destabilising terrain, as QS calculated for the weight acceptance phase would not accurately represent QSf and QSe. A two-spring system whereby one spring is set to QSf and the other is set to QSe and only function in their corresponding stages of the gait cycle may present a better alternative.

*R*2 values for QSf and QSe were higher during the control condition as opposed to the destabilising terrain condition indicating that the relationship between the applied moment and knee excursion was more linear when walking over level ground. Specific values for control were in-line with those asserted in previous literature [23,55]. It has been previously determined that the linearity of spring-like behaviour changes with gait speed [55] and therefore could be responsible for *R*2 differences between conditions. Unfortunately, such results prevent conclusions being drawn asserting that differences in linearity is solely due to terrain perturbations.

A significant difference was found in gait speed between the even and destabilising surface trials, concurring with the findings of Menant et al. [56]. Mean data (Table 1) highlight that the Fr was 0.05 lower when walking across destabilising terrain which calculates to a difference on 0.74 metres per second for an individual with a leg length of 0.91 m. This supports the finding regarding greater QS over destabilising terrain as it has previously been established that slower gait speeds facilitate greater compliance at the knee [41,57].

A greater stride width was observed on a destabilising terrain conditions (Table 1), agreeing with findings of Menant et al. [56]. Strides were 17.1% wider when ambulating on the destabilising surface, due to the unfamiliarity of the uneven surface thought to cause balance issues. Specifically, increased stride width is associated with a gait adaptation that attempts to improve walking stability in the frontal plane [56,58,59].

Although these findings regarding gait speed, stride length and stride width offer insight into gait adaptation mechanisms when ambulating over a destabilising terrain, they do not allow for judgements to be made as to whether QS predictive models based upon both body and gait parameters [16] are capable of accurate results as observed changes in gait speed would be considered in the model.

Data was presented on kinematics and kinetics of the knee joint in the sagittal plane only, as it was those movements that are related to the stiffness of the knee and the vertical motion of the participant’s centre of mass, which were required to answer our research question. Movement is the transverse and coronal planes may be of interest if a realistic range of motion was to be incorporated in a prosthetic, but were beyond the scope of the research question answered here. Lower limb motions in the transverse and coronal planes mainly emanate from femoral motion at an intact hip joint, and would be an interesting focus for future work.

Further work is required to understand how the QS of the ankle and hip joints change when walking on a destabilising terrain. This would offer insight as to whether estimation models focused on those joints [60,61] would have to be adapted between even and destabilising terrain.

***5.0 Conclusion***

This investigation has highlighted the importance of adapting current predictive, stature-based models of QS to align with the requirements of walking across a destabilising terrain. A primary finding was that QS was significantly greater when walking across a destabilising terrain in comparison to even ground. In addition, QSf and QSe when walking across even ground were seen to resemble spring-like behaviour, supporting the previous literature [16]. When walking across destabilising terrain, however, QSf was seen to be much higher than QSe indicating that the use a single torsional spring to tune a TFP would not be suitable.

Biomedical engineers may utilise these findings to develop a TFP that is capable of accurately emulating the human gait cycle across both even and destabilising terrains. Additionally, findings are also suitable for implementation in to the design and tuning of lower limb wearable exoskeletons and are not restricted to TFP design alone.

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1. Transfemoral prosthesis (TFP) [↑](#endnote-ref-1)
2. Quasi-stiffness (QS) [↑](#endnote-ref-2)
3. Newton-metres per radian (Nm·rad­-1) [↑](#endnote-ref-3)
4. Metres (m) [↑](#endnote-ref-4)
5. Kilograms per metres squared (kg·m-2) [↑](#endnote-ref-5)
6. Froude number (Fr) [↑](#endnote-ref-6)
7. Centimetres (cm) [↑](#endnote-ref-7)
8. Centre of pressure (CoP) [↑](#endnote-ref-8)
9. Quasi-stiffness of the flexion sub-phase of the weight acceptance phase (QSf) [↑](#endnote-ref-9)
10. Quasi-stiffness of the extension sub-phase of the weight acceptance phase (QSe) [↑](#endnote-ref-10)
11. Coefficient of determination (*R*2) [↑](#endnote-ref-11)