VA RESEARCH & DEVELOPMENT

JRRD
Journal of Rehabilitation Research & Development
Vol. 36 No. 3, July 1999

Prosthetic loading during kneeling of persons with transfemoral amputation

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Abstract--Observations in the field of lower limb prosthetic rehabilitation have shown that several transfemoral prostheses show signs of wear on some components of the knee unit. This is thought to be a result of severe loading developed during activities associated with kneeling. Some prostheses may have failed due to repetitive action of such loading. In order to determine the nature and magnitude of the loads developed during kneeling by persons with transfemoral amputation, and to investigate the influence of various prosthetic parameters, an analysis of the results of 162 tests in prosthetic knee hyperflexion was undertaken. The services of four males with amputation were enlisted. The measurements involved simultaneous use of two Kistler force platforms, a six-channel strain gauge transducer mounted on the prosthetic shank, and a data acquisition system. The critical loads for this configuration were found to be the shear force on the knee hinge, the shear force imposed by the knee chassis on the shin, and the bending moment tending to hyperflex the knee. These loads ranged from 0.6 to 6.2 kN, 0.9 to 6.7 kN, and from 18.3 to 155.7 Nm, respectively. To achieve a comfortable kneeling position, some prostheses permit foot rotation about the pylon axis of 90° to allow the shank to be approximately parallel to the ground. Tests were also conducted with the prostheses in this configuration and the most influential prosthetic parameter was found to be the external rotation of the foot (toe-out angle). During kneeling, it was found that the loading was dependent upon the position of the torso relative to the prosthesis, but loads were much higher than those developed during level walking.

Key words: *kneeling, prosthetic loading, standards, testing, transfemoral prosthetics.*

INTRODUCTION

Kneeling is an activity that is performed daily by millions of people for vocational, cultural, or religious reasons. It is an activity undertaken by persons with amputation and, anecdotally at least, the loads imposed on a prosthetic knee during kneeling have been held responsible for premature wear of the knee components and unexpected failures. However, to the authors' knowledge, no previous attempt has been made to determine the loading on a prosthesis during kneeling. Therefore, this investigation was undertaken to increase knowledge of the mechanical behavior of transfemoral prostheses during kneeling and avoid the occurrence of damage that might impair function, with potentially hazardous consequences.

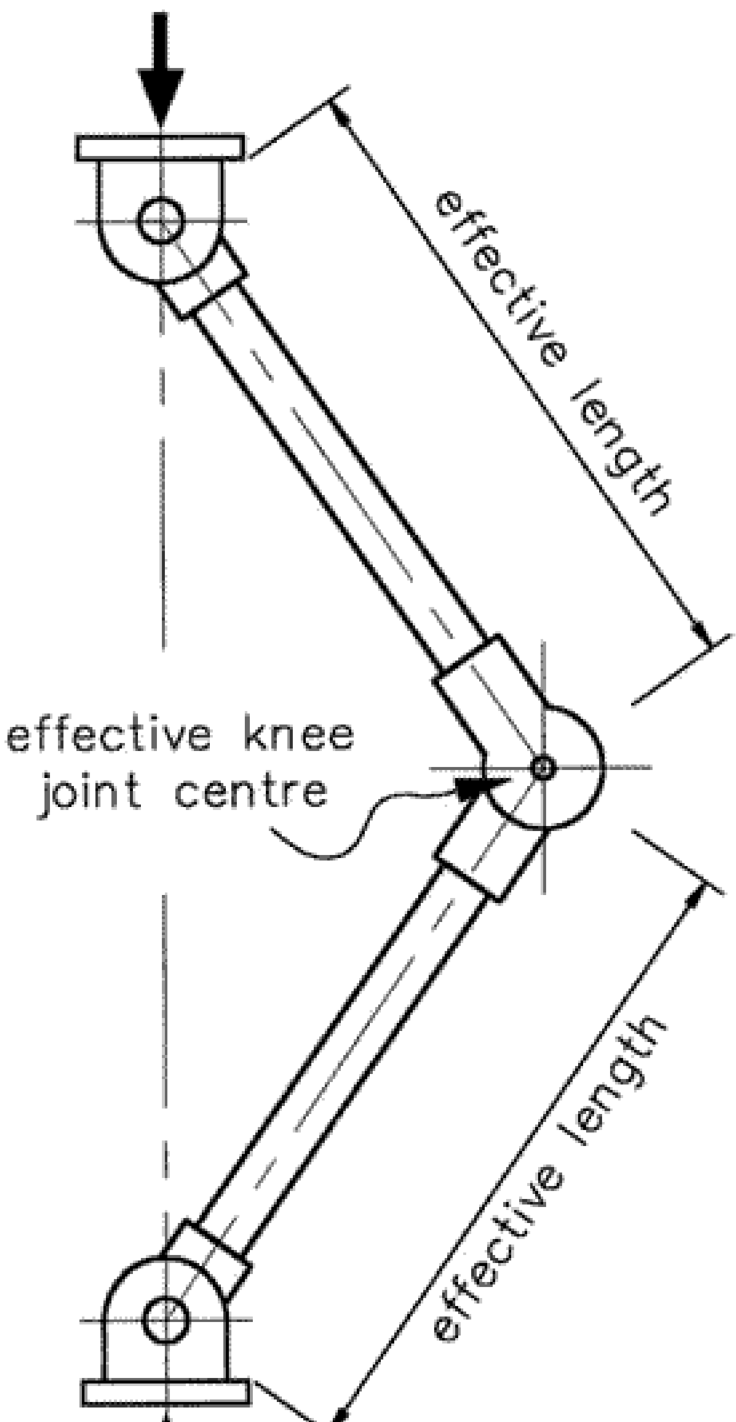
Historically, the development of lower limb prostheses took place for many years without any form of structural testing other than field use by the person with amputation. For this purpose, the Department of Veterans Affairs in the United States maintained a population of persons with amputation of above average body mass and higher than normal activity level. Before the advent of modular prostheses, each prosthesis was individually fabricated for each user; thus, the results of any mechanical testing procedure obtained from one prosthesis could not be applied directly to others. However, with the introduction of modular lower limb prosthetic systems and the application of engineering principles to prosthesis design, it became desirable to establish the safety of the product prior to commencing mass production.

At the final stages of the design process, it was possible to verify the stress calculations by applying simulated loading conditions in a test-rig. Following the ISPO Philadelphia meeting in 1977 (1), the United Kingdom Department of Health and Social Security started a program of structural testing of all new lower limb prostheses, and, in parallel, it introduced a system of records to monitor all defects and adverse events occurring in the field.

The advent of the Modular Assembly Prosthesis (M.A.P; Chas. A. Blatchford & Sons, Ltd.) saw the introduction of a mechanical hyperflexion stop that limited the knee flexion angle of the prosthesis during kneeling. Ongoing collaboration between Chas. A. Blatchford & Sons, Ltd., the Department of Health, and the Bioengineering Unit, University of Strathclyde resulted in evolution of a method for structural testing being established and the application of the knee-flexion stop test was improved by reviewing the applied test load. It was ultimately included among other supplementary tests within ISO 10328 (2). It was agreed to revise this between 1998 and 1999; however, two problem areas remained to be addressed:

1. As the test method and values evolved around specific limb systems, it was found that due to the variation in the maximum knee flexion angle from one prosthetic system to another, there was a possibility that a system with a large flexion angle might be subjected to a higher bending moment in real life than a system with a smaller maximum knee flexion angle.
2. The test load was based on records of failed prostheses, and there were no data available on the value of the load actually applied during kneeling by the prosthesis wearer.

To solve these two problems, the UK members of Working Group 3 of ISO Technical Committee 168 (Prosthetics and Orthotics - Testing) proposed a test method that was independent of the knee flexion angle. The test method and performance requirements are stated in ISO 10328 parts 5 and 6 (2). The line of action of the applied load of 1750 N passes through a point in the thigh 400 mm from the knee joint center and through a point in the shank 400 mm from the knee joint center (**Figure 1**). This means that the knee moment applied to a system with a maximum flexion angle of 120° is 606 Nm and for a system with 160° of knee flexion angle the moment is 689 Nm, due to the increased moment arm. Both of these values are above 600 Nm, which from UK field experience, had been considered a level that would assure safety but would not lead to overdesign. In order to address these problems and ascertain the actual load levels determined during kneeling, a project was initiated at the Bioengineering Unit of the University of Strathclyde, Glasgow.



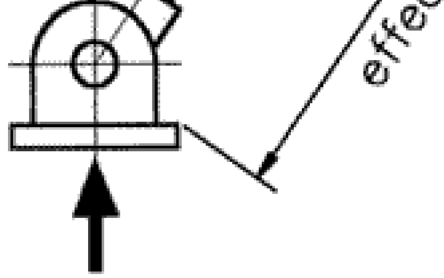


Figure 1.
Recommended configuration for mechanical testing of knee flexion stops; effective length = 400 mm. Adapted from ISO 10328 Part 5 (1996).

METHODS

Subjects and Prostheses

The services of four established males with amputation were recruited and they gave written informed consent. **Table 1** exhibits data related to these four subjects, referred to as subjects #1, #2, #3, and #4. Total contact quadrilateral sockets were prepared for all subjects and mounted on Endolite prostheses with uniaxial knee, manual lock, and ESK stance flex mechanism (Chas. A. Blatchford and Sons, Ltd., Basingstoke, UK).

Table 1.
Physical details of test subjects.

Subject No.	Side*	Activity Level**	Mass+	Height++	A mm	B mm	C mm
1	right	37	100.0	1.88	360	500	290
2	left	30	71.4	1.80	320	510	140
3	right	41	73.3	1.87	385	515	170
4	right	12	72.1	1.74	325	500	260

* Side of amputation; **Activity level after Day, 1981 (3); + Mass = body mass including prosthesis, in kg; ++ Height in m, including shoes; A = distance from ischial tuberosity to knee center; B = distance from knee center to bottom of heel; C = residual limb length.

Equipment and Instrumentation

The equipment used in these tests was as follows (see **Figure 2**):

- two Kistler force platforms (Kistler AG, Winterthur, Switzerland), installed in the gait analysis laboratory of the Bioengineering Unit of the University of Strathclyde
- the six-channel Strathclyde pylon transducer (4) incorporated at the distal end of the shin tube of the prosthesis and connected to specially built strain gauge amplifiers
- A microVax computer (Digital Equipment Corporation) connected to the analog-to-digital

converter (ADC) to which all other devices were also connected.

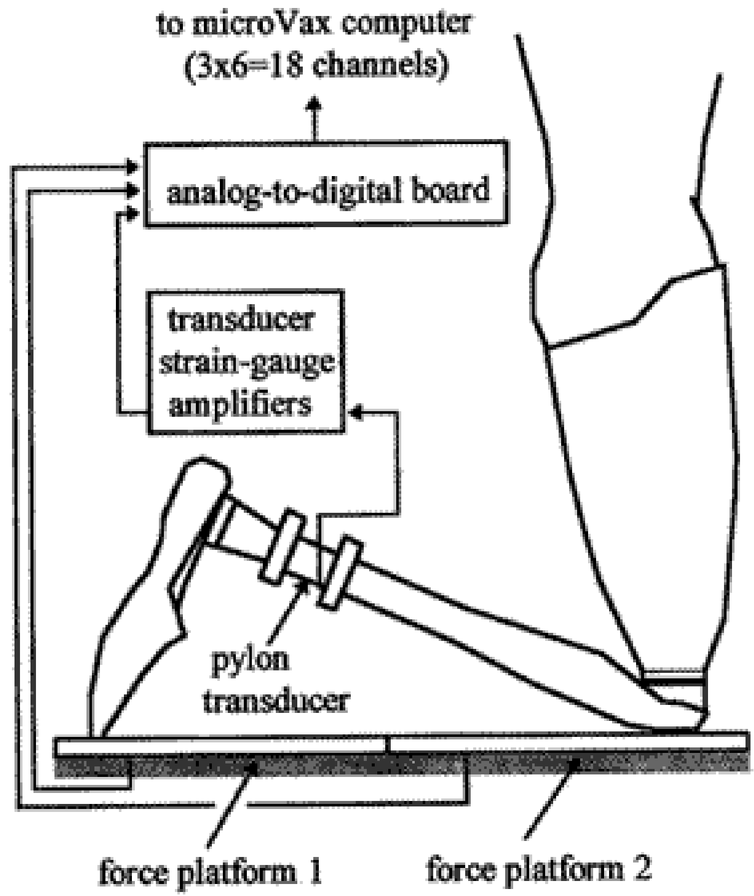


Figure 2.
The experimental set-up for the kneeling tests.

The output signals of each force platform correspond to the six components of the ground reaction loads (three force and three moment components). The six output signals of the pylon transducer correspond to the three components of the force and the three moment components transmitted by the prosthesis at its location in the shin tube. Thus, a total of 18 channels was recorded by data acquisition governed through the software of the microVax terminal. Based on the records of the timing of preliminary testing, it was decided to adopt a data acquisition duration of 30 seconds. A sampling frequency of 50 Hz was considered adequate.

Mathematical Formulation and Mechanics

Mechanical failure (deformation, crack growth, fatigue) of prosthetic components due to kneeling corresponds to the internal structural loads developed, and these depend on the particular design of each unit and the associated lever system.

During preliminary tests, it was noted that the knee axis during kneeling was approximately horizontal. Thus, it was possible to simplify the problem by considering the analysis of the configuration in the vertical plane only. The prosthetic kneeling configuration described in **Figure 3** was, therefore, adopted for the geometric description, mathematical formulation, and mechanical solution of the problem. Since kneeling is a quasi-static situation, the tests were conducted and analyzed without consideration of dynamic effects on components between the force-measuring equipment and the knee axis. The test setup takes into account the major geometric prosthetic parameters (see **Figure 3a**), which are the distances between the various landmarks, and can be measured on a bench before testing.

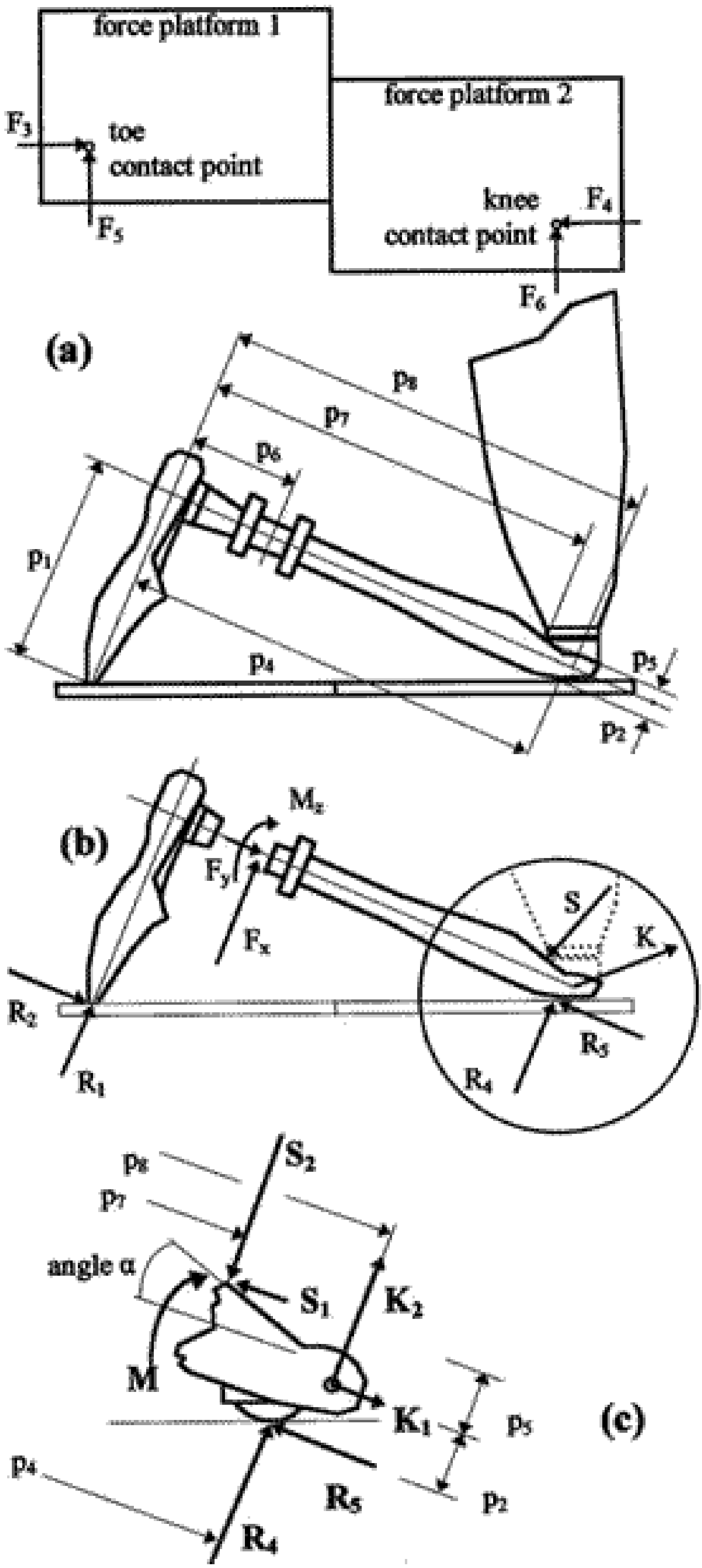


Figure 3.

The kneeling configuration under study: (a) view of the force platforms and the prosthesis in the kneeling position; (b) loading on the prosthesis in the kneeling configuration; and, (c) detail of loading on the knee unit.

Figures 3b and **3c** exhibit the loads developed and monitored during kneeling on an Endolite prosthesis. Using the geometric parameters of the force platform and prosthesis setup (**Figure 3a**), the components F_1 to F_6 of the ground reaction forces, expressed with respect to the force platform frames of reference, are converted to components R_1 to R_6, expressed with respect to a frame of reference parallel to the one associated with the prosthesis and the pylon transducer. The load components R_1, R_2, R_4, R_5 and internal pylon transducer reaction components F_x, F_y, and M_z can be used to evaluate the unknown kneeling loading, which is described by the following quantities: the shear force K applied by the knee axle (fulcrum) across the wall of the shin, and the shear force S and bending moment M applied, as a result of the "nutcracker" effect, along the cross section, which is defined by the posterior side of the knee chassis of the shin (**Figure 3c**).

Load components K, S, and M can be evaluated by using two different free-body diagrams: 1) that of the prosthesis between the two force platforms, ignoring the pylon transducer (referred to as Method A) and 2) that of the prosthetic part between the pylon transducer and force platform 2, ignoring force platform 1 (referred to as Method B).

Method A

For **Method A** by equilibrium of:

forces parallel to the transverse axis,

$$\mathbf{-R_1 + S_2 - R_4 - K_2 = 0} \quad \textbf{[1]}$$

forces parallel to the longitudinal axis,

$$\mathbf{R_2 - S_1 - R_5 + K_1 = 0} \quad \textbf{[2]}$$

moments about the axis through the prosthetic toes,

$$\mathbf{R_4p_4 + R_5\,(p_1\text{-}p_2) - K_1p_1 + K_2p_8 + S_1\,(p_1\text{+}p_5) - S_2p_7 = 0} \quad \textbf{[3]}$$

and

$$\mathbf{M = R_1p_7 - R_22p_1} \quad \textbf{[4]}$$

Method B

For **Method B** by equilibrium of:

forces parallel to the transverse axis,

$$\mathbf{-F_x - R_4 - K_2 + S_2 = 0} \quad \textbf{[5]}$$

forces parallel to the longitudinal axis,

$$\mathbf{F_y - S_1 - R_5 + K_1 = 0} \quad \textbf{[6]}$$

moment about the transverse axis through the pylon transducer center,

$$R_4 (p_4 - p_6) - R_5 p_2 + K_2 (p_8 - p_6) + S_1 p_5 - S_2 (p_7 - p_6) - M_z = 0 \quad [7]$$

and

$$M = F_x (p_7 - p_6) + M_z \quad [8]$$

In both methods the two components of the shear force S are related by:

$$S_1/S_2 = \tan \alpha \quad [9]$$

where α is the known angle of the inclination of the posterior wall of the shin to the axis of the prosthesis, shown in **Figure 3c.**

Furthermore, the use of the two force platforms gives the opportunity to measure the amount w percent of the subject's body weight W carried by the prosthetic side during kneeling and the corresponding lever arm of the body weight vector, with respect to the prosthetic knee expressed as d percent of the base distance D measured on the ground between the knee and foot (**Figure 4**):

$$w = [(F_1 + F_2) / W] \cdot 100 \quad [10]$$

$$d = [F_1 / (F_1 + F_2)] \cdot 100 \quad [11]$$

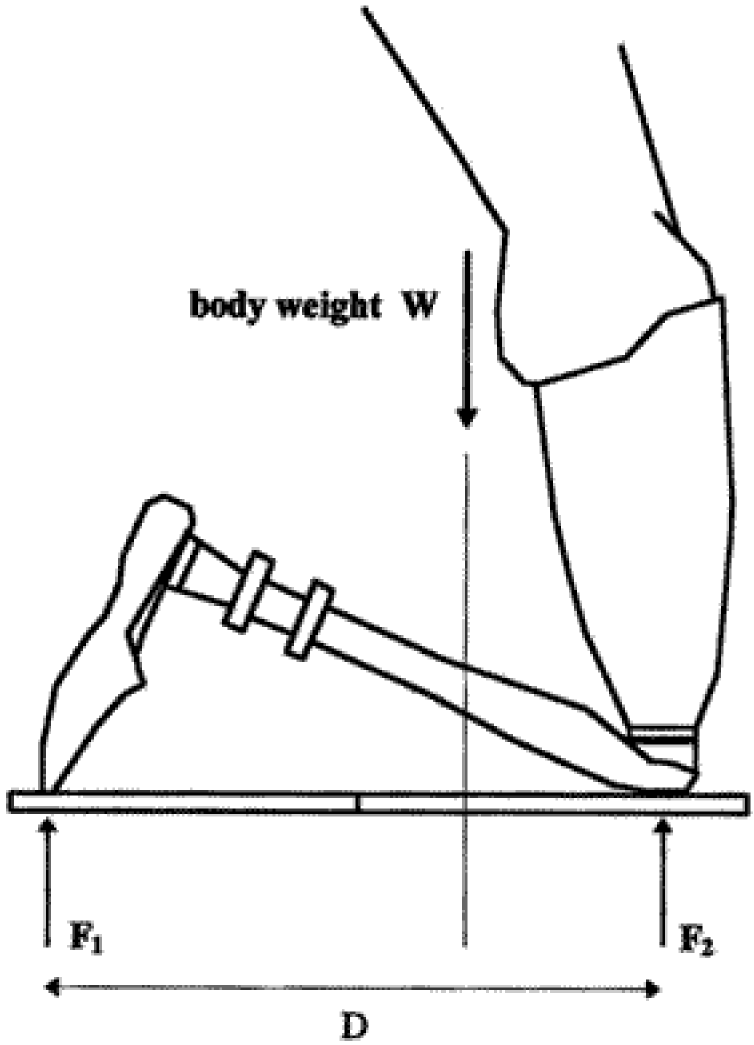


Figure 4.
Representation of body weight direction with respect to force platforms and vertical components of ground reaction forces.

Processing the Data

Dedicated software was developed for processing and analysis of the acquired data. The software was designed to first convert raw data to meaningful mechanical quantities (forces and moments), using the calibration factors of the force platforms and of the pylon transducer, and then to select the data set corresponding to the instant at which the subject leaned his torso back furthest. This instant was identified by the maximum value of load component F_1 on force platform 1, below the prosthetic foot (see **Figure 4**). The set of data, produced by this kneeling configuration, was used to solve the equations, evaluate kneeling loads K, S, and M, and calculate percentages w and d.

Testing and Calculation Procedures

To determine the maximum static loads developed within the prosthetic knee unit during kneeling and to investigate the influence of certain parameters (such as limb alignment, type of prosthetic foot, damping properties of the knee stanceflex mechanism, and changes to the toe-out angle) on the kneeling loads developed, three series of tests were performed. A comparison between calculation Methods A and B was also undertaken.

Series 1

Twenty-seven tests involving subject #1 were performed during which the socket was first held in a neutral alignment position and loads were measured with a multiflex foot and then with a SACH foot attached to the prosthesis. Data were recorded with the foot set at zero toe-out angle and at 90° toe-out angle. Maintaining the SACH foot on the prosthesis, the socket alignment was then set in a maximally flexed position and shifted fully posteriorly and then fully anteriorly, again with the foot at zero and at 90° toe-out angles. Additionally, all of these tests were performed with a rubber stanceflex bumper *in situ* and then with a steel block to prevent socket flexion from occurring.

Series 2

A further set of 15 tests were performed on subject #1 with the prosthesis optimally aligned to the subject's satisfaction with the rubber stanceflex bumper *in situ.* Both calculation methods, A and B, were used to provide the results from these sets of 42 tests. While the results were comparable in most cases, it was found that those obtained from Method B yielded lower values than Method A. This is thought to arise from minor deviations of the knee axis from the horizontal. Since the pylon transducer frame of reference is directly related to the prosthesis frame of reference, the results obtained using Method B were considered more reliable and only this method was adopted for the calculation of results in the third series of tests.

Series 3

This third phase consisted of 120 tests involving subjects #2, #3, and #4 with their prostheses set to an optimum alignment position. This number of tests provided statistical validity to the ranges of the calculated kneeling loads. In order to record data expressing the datum of the output signals in each one of the 18 channels (i.e., signals under zero load), the subjects were asked to stand away from the platforms and lift their prostheses off the ground. One single brief period of data acquisition supplied data corresponding to zero load for all 18 channels. Actual tests could then take place provided that all instruments were reset immediately prior to the activity. This was necessary in order to minimize the effect of drift particularly for the Kistler force platform signals, which are transmitted via charge amplifiers. As shown in **Figure 2,** during actual tests the subjects were asked to kneel against force platform 2 and position the prosthetic foot with the toes against force platform 1, leaning their torso as far back as possible, but maintaining a sense of security and stability. It was at this final stage of each trial that actual data acquisition was performed in order to avoid creation of large files with unnecessary data.

RESULTS

Tables 2 and **3** exhibit the values of shear forces K and S, bending moment M, and percentages w and d calculated by the software for the tests of the first and second series, respectively. These two tables present the results obtained using both calculation Methods A and B.

Table 2.

Results calculated from the first test series on subject #1 only.

Test No.	Parameters@ Align* T	Sh	Foot type +	T-O angle	Force K (kN) Method A	B	Force S (kN) Method A	B	Moment M (Nm) Method A	B	w (%)	d (%)
1&4	n	n	M	0°	2.3 (3.3)	2.2 (3.3)	2.8 (3.7)	2.6 (3.7)	59.7 (85.0)	56.1 (85.0)	58.7	24.9
2&5	n	n	M	0°	2.5 (2.9)	2.3 (2.5)	2.9 (3.1)	2.7 (2.9)	62.1 (68.8	57.7 (64.6)	52.3	25.9
3&6	n	n	M	0°	2.5 (4.3)	2.4 (4.2)	2.9 (4.7)	2.8 (4.6)	61.9 (107.0)	59.2 (104.4)	56.0	25.9
7	n	n	M	90°	5.9	5.6	6.5	6.2	145.0	136.8	65.9	49.5
8	n	n	M	90°	4.8	4.5	5.4	5.0	118.4	110.9	56.0	47.5
9	n	n	M	90°	4.4	4.2	5.0	4.2	109.2	103.0	58.7	42.1
10	n	n	S	0°	2.4	2.2	2.8	2.6	59.9	55.9	53.2	24.9
11	n	n	S	0°	2.1	1.9	2.5	2.3	54.7	48.7	49.8	22.6
12	n	n	S	0°	2.7	2.5	3.2	2.9	68.3	63.1	59.3	28.4
13	n	n	S	90°	3.6	3.5	4.1	4.0	88.1	84.7	55.0	40.7
14	n	n	S	90°	3.6	3.5	4.2	4.1	88.7	85.7	58.6	40.9
15	n	n	S	90°	--	--	--	--	--	--	--	--
16	x	b	S	0°	3.1	2.9	3.5	3.3	79.8	74.6	52.2	33.2
17	x	b	S	0°	3.1	2.9	3.6	3.3	79.1	73.5	54.2	32.9
18	x	b	S	0°	3.7	3.5	4.2	3.9	94.7	88.2	54.5	39.4
19	x	b	S	90°	3.1	2.8	3.6	3.3	76.9	74.6	46.8	35.7
20	x	b	S	90°	3.2	2.9	3.6	3.3	82.2	73.5	40.6	38.4
21	x	b	S	90°	3.8	3.6	4.3	4.1	95.8	88.2	52.7	44.4
22	x	f	S	0°	3.6	3.4	4.1	3.9	91.8	86.3	61.2	38.2
23	x	f	S	0°	3.1	2.9	3.6	3.4	80.9	75.4	55.0	33.7
24	x	f	S	0°	3.6	3.3	4.1	3.8	92.2	86.1	58.1	38.4
25	x	f	S	90°	4.6	4.6	5.1	5.0	115.4	113.6	45.9	53.7
26	x	f	S	90°	5.3	5.0	5.8	5.5	133.2	126.6	54.6	62.0

27	x	f	S	90°	4.4	4.0	4.9	4.5	109.3	99.7	50.0	50.2

@ Prosthetic parameters; *Anterioposterior alignment: T=tilt, Sh=shift, n=neutral, x=fully flexed, b=back, f=fore; +Foot Type: M=Multiflex foot, S=SACH foot; T-O=toe-out; data acquisition for Test 15 was unsuccessful.

Table 3.

Results from the second test series on subject #1 only.

	Force K (kN)		**Force S (kN)**		**Moment M (Nm)**			
Test	**Method**		**Method**		**Method**		**w**	**d**
No.	**A**	**B**	**A**	**B**	**A**	**B**	**(%)**	**(%)**
28	5.3	5.3	5.8	5.7	133.7	133.7	51.6	51.4
29	4.6	4.5	5.0	4.9	115.5	112.4	47.6	48.4
30	5.9	5.9	6.4	6.3	150.1	147.8	55.9	53.4
31	4.8	4.8	5.2	5.2	121.5	120.4	47.9	50.3
32	5.7	5.5	6.2	6.0	144.1	139.5	51.2	55.9
33	4.7	4.5	5.1	4.9	117.4	111.7	49.7	47.1
34	--	--	--	--	--	--	--	--
35	4.7	4.3	5.2	4.7	119.8	108.2	50.9	46.7
36	5.1	4.7	5.6	5.2	130.6	120.6	57.3	45.1
37	5.3	4.9	5.8	5.4	134.5	124.6	58.7	45.4
38	5.2	4.8	5.7	5.3	130.8	122.0	57.1	45.4
39	5.0	4.6	5.5	5.1	127.8	117.3	55.9	45.3
40	6.1	5.8	6.6	6.2	154.2	146.2	56.3	54.2
41	5.7	5.3	6.2	5.8	144.9	135.0	53.3	53.9
42	6.6	6.2	7.1	6.7	165.6	155.7	59.3	55.4

Method A: using data from force platforms 1 and 2; Method B: using data from pylon transducer and force platform 2; data acquisition for Test 34 was unsuccessful.

It can be appreciated that the values of the load components K, S, and M calculated using Method B were consistently lower than those calculated using Method A, but the maxima and minima of these values were obtained, for both methods, during the same tests. The maxima and minima of quantities w and d however, were not always obtained during the same tests.

For the tests of the first series and referring to calculation Method B, it can be noted that shear force K varied between 1.9 and 5.6 kN, shear force S varied between 2.3 and 6.2 kN, and bending moment M varied between 48.7 and 136.8 Nm. For the tests of the second series and referring to calculation Method B again, it can be noted that shear force K varied between 4.3 and 6.2 kN, shear force S varied between 4.7 and 6.7 kN, and bending moment M varied between 108.2 and 155.7 Nm.

For the first series of tests, percentage w of the body weight, applied on the prosthetic side, generally varied between 40.6 and 65.9 percent and the lever arm expressed as a percentage of the knee-to-foot distance varied between 22.6 and 62.0 percent. For the second series of tests, percentage w varied between 47.6 and 59.3 percent and lever arm d varied between 45.1 and 55.9 percent.

Table 4 exhibits a global summary of statistics of the quantities under study as derived using Method B, for all subjects with their preferred settings (i.e., tests of second and third series).

Table 4.

Results from the second and third test series analyzed using Method B only.

Subject	No. *	Force K (kN)		Force S (kN)		Moment M (Nm)		w (%)		d (%)	
		R+	M++	R	M	R	M	R	M	R	M
#1	15	4.3-6.2	5.1 ±0.6	4.7-6.7	5.5 ±0.6	108.2-155.7	128.2 ±14.9	47.6-59.3	53.8 ±4.0	45.1-55.9	49.9 ±4.1
#2	40	0.6-2.1	1.2 ±0.4	0.9-2.5	1.6 ±0.4	18.3-53.1	32.5 ±9.0	58.5-69.6	62.9 ±3.1	13.9-32.0	21.8 ±4.3
#3	40	3.3-5.3	4.1 ±0.5	3.7-5.7	4.5 ±0.5	83.3-129.5	101.3 ±11.4	54.8-69.2	60.6 ±4.0	47.9-69.3	56.7 ±5.6
#4	40	0.6-2.6	1.6 ±0.5	1.0-3.0	2.0 ±0.5	19.4-66.7	42.0 ±12.1	53.9-67.0	60.1 ±2.7	14.9-39.8	26.7 ±6.4

* No. = number of tests; + R = range; ++ M = mean ± standard deviation; w = percentage of body weight applied to the prosthesis; d = percentage of prosthesis load transmitted at the foot.

DISCUSSION

The main task was to establish values for the high static loads developed within the knee unit of transfemoral prostheses during kneeling and investigate the influence of various parameters on the magnitude of these loads.

Prior to testing, the anticipated result was that the loads would be increased for any test alignment that increased the posterior translation of the subject's center of gravity (i.e., for any configuration that would allow the subject to lean further back). Such configurations were thought to involve maximum socket flexion, maximum posterior socket shift, toe-out rotation angle of the foot, and perhaps a more compliant stance-flex mechanism. Thus, a configuration that would involve all of these properties simultaneously was expected to result in the most severe kneeling loads.

However, the above considerations proved to explain only half the story, leaving the rest to the subject's own initiative during testing. As proved by the results reported for the first testing series (**Table 2**), the major determinants for the magnitude of the kneeling loads were not the objective alignment parameters mentioned above but the quantities w and d, which were subjectively imposed by subject #1 as a result of his sense of stability and proprioception. Although the prosthetic configuration adopted during tests numbered 19 to 21 appeared likely to induce the highest loading, this did not occur. Because of the sense of insecurity caused by this configuration, subject #1 relieved his prosthetic side (relatively low w ranging from 40.6 to 52.7 percent), relying more on his sound side and simultaneously translated his center of gravity forward (relatively low d ranging from 35.7 to 44.4 percent), using his hip musculature.

Severe kneeling loads were developed by this heavily built subject at the beginning of the first testing series (**Table 2**), when the subject was not yet familiar with the procedure; thus, the determining factor was the alignment configuration (as in tests numbered 7 to 9, where a drastic 90° toe-out rotation was imposed). Severe kneeling loads were also developed during the second testing series, as in test number 42 (**Table 3**), where subject #1, much more familiar with the procedure and with his preferred alignment, voluntarily applied more weight on the prosthetic side (w=59.3 percent) and leaned safely back with confidence (d=55.4 percent). In this test, loads K, S, and M reached their maxima 6.2 N, 6.7 N, and 155.7 Nm, respectively. The normal safety factors incorporated in the test values in the standard are for proof load values to be about twice the value in normal activities and failure load about 4 times this value. Thus, the ultimate test load specified in the standard, which is 600 Nm, is in accordance with this practice.

Referring to **Table 4,** it can be appreciated that the loads developed during the kneeling tests of subjects 2, 3, and 4 were also high, although their maxima are lower compared to those of the heavily built subject #1. However, it is important to repeat that the magnitude of these loads is determined not only by the body mass but also by the voluntary application of weight and positioning of the torso.

Among the prosthetic parameters, toe-out angle (external rotation of the foot) was found to be more important in the development of high kneeling loads. Any rotation of the foot at the kneeling position (in this case, 90° toe-out) effectively decreases the distance of the prosthetic heel from the ground and consequently the proximal part of the socket is positioned further posteriorly. This situation increases the lever arm of the applied weight (increased "nutcracker" effect) and consequently the loads developed within the knee unit. This conclusion has an implication that installation of an ankle rotary mechanism will increase loading on the knee. However, this has to be offset against the increased security and comfort to the person with amputation as is stated to happen for subjects with transtibial amputation (5).

It should be noted that the results obtained by these tests were considered in the formulation of the International Standards ISO 10328, Parts 5 and 6 (2).

CONCLUSION

Independent of any prosthetic parameters, the maximum loads within the knee unit during kneeling are developed when the person with transfemoral amputation applies the greatest load on his prosthetic side and leans back furthest. Under these conditions, knee units are mechanically challenged and, depending on their design, some components could be subject to heavy loads, although in most cases only infrequently. Shear forces and bending moments reported here markedly exceeded normal walking loads, which are generally used for the structural testing of prosthetic systems.

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