JOINT LOAD CONSIDERATIONS IN TOTAL KNEE REPLACEMENT

MARKUS S. KUSTER, GRAEME A. WOOD, GWIDON W. STACHOWIAK, ANDRÉ GÄCHTER

From the University of Western Australia, Australia

Estimates of knee joint loadings were calculated for 12 normal subjects from kinematic and kinetic measures obtained during both level and downhill walking. The maximum tibiofemoral compressive force reached an average load of 3.9 times body-weight (BW) for level walking and 8 times BW for downhill walking, in each instance during the early stance phase. Muscle forces contributed 80% of the maximum bone-on-bone force during downhill walking and 70% during level walking whereas the ground reaction forces contributed only 20% and 30% respectively.

Most total knee designs provide a tibiofemoral contact area of 100 to 300 mm². The yield point of these polyethylene inlays will therefore be exceeded with each step during downhill walking. Future evaluation of total knee designs should be based on a tibiofemoral joint load of 3.5 times BW at 20° knee flexion, 8 times BW at 40° and 6 times BW at 60°.

J Bone Joint Surg [Br] 1997;79-B:109-13.

Received 6 June 1996; Accepted 24 September 1996

Several recent studies have reported severe wear of polyethylene tibial components.¹⁻⁴ The long-term problems associated with joint wear debris, such as loosening and infection, are well known. Wear is dependent on a number of factors including contact area, load, material properties, thickness of the polyethylene inlay and the length of time that the component has been implanted.^{1,5} The most destructive wear process is fatigue, which occurs through

repeated high loads and cyclic stressing. Load is dependent both on physical activity and on body-weight.

The increasing long-term successes being achieved with total knee replacement means that younger, and consequently more active, patients are being treated. This places an increased mechanical demand on the prosthesis which exceeds the design limits of many of the currently used devices.

The moments and forces about the knee vary substantially for different daily activities.⁶ Biomechanical studies of knee joint loading have consistently estimated maximum joint compressive forces to be about 4 to 4.5 times body-weight during daily activities.⁷ This range of values has become a design criterion for most currently used knee prostheses, but recent studies have indicated that loadings can be much higher even during level walking.^{9,10} This finding is consistent with the increasing incidence of reports of severe wear in joint replacements.^{1,2,4,11-14} We present quantitative joint load data and suggest new criteria for use in the biomechanical evaluation of total knee prostheses.

MATERIALS AND METHODS

We obtained estimates of knee joint loading for 12 normal subjects (6 male and 6 female) ranging in age from 23 to 37 years (mean 27.9), in height from 158 to 187 cm (mean 171) and in weight from 49 to 90 kg (mean 70.8). Reflective markers were located superficial to the 5th metatarsophalangeal, ankle, knee and hip joints. Spatial trajectories were recorded using a video-based motion analysis system with two cameras sampling at 60 Hz (APAS, Ariel Dynamics, Inc, Trabuco Canyon, California) whilst the subjects walked across a level floor and down a purpose-built ramp of 19% gradient. Ground reaction force data were simultaneously obtained from a Kistler force platform (Type 9281B, Winterthur, Switzerland). Ground reaction forces during downhill walking were measured using an aluminium plate bolted to the force platform. Step frequency for both downhill and level walking was controlled by means of a metronome set at 120 steps/min.

The marker trajectories in the sagittal plane were smoothed using a Butterworth 4th-order low-pass digital filter with a cut off frequency of 7 Hz prior to the derivation of segmental orientations and centre of mass locations. Time derivatives of these measures were then calculated by finite differences. Finally, planar joint reaction forces and net

M. S. Kuster, MD, Orthopaedic Surgeon
A. Gächter, Professor and Chairman
Klinik für Orthopädische Chirurgie, Kantonsspital, 9007 St Gallen, Switzerland.

G. A. Wood, PhD, Associate Professor
Neuromuscular Performance Laboratory, Department of Human Movement
G. W. Stachowiak, PhD, Associated Professor
Tribology Laboratory, Department of Mechanical and Materials Engineering
The University of Western Australia, Nedlands, Australia 6907, Australia.

Correspondence should be sent to Dr M. S. Kuster.

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0301-620X/97/16978 \$2.00

joint moments at the ankle, knee and hip were estimated from these kinematic data and force platform measures using standard inverse dynamics procedures and anthropometric values from Winter.¹⁵

A knee joint model, previously described by Nisell,¹⁶ was then used to calculate the tibiofemoral (bone-on-bone) force from the mean joint reaction forces and knee extensor moments which had been derived from the inverse dynamic analysis. The total uncertainty based on the standard error of the mean of the peak bone-on-force was calculated as outlined by Campion.¹⁷ Full details of the instrumentation used can be found in the authors' previously published work.^{18,19}

RESULTS

The highest knee joint loadings occurred during downhill walking. The peak joint moments occurred at $41 \pm 6^\circ$ knee flexion and were 2.75 ± 0.5 Nm/kg for females and 2.70 ± 0.7 Nm/kg for males. The vertical joint reaction forces were 15.2 ± 1.6 N/kg for females and 15.5 ± 1.9 N/kg for males. The kinetic and kinematic data for males and females were not significantly different ($p > 0.85$). The joint model used in this investigation defined the lever arm of the quadriceps muscles of female subjects as being significantly smaller than that of male subjects. Thus the

actual tibiofemoral model predictions were consistently smaller for male subjects. The peak tibiofemoral force for male subjects was 7 times BW during downhill walking, whereas it reached 8 times BW for female subjects. Values obtained for level walking (at 20° knee flexion) were approximately 50% of those for downhill walking giving values of 3.4 times BW for male and 3.9 times BW for female subjects. The standard error of the mean in the prediction of peak force was calculated to be 13% for level and downhill walking. The mean female knee joint compressive forces for the duration of the support phase during downhill walking and level walking are shown in Figure 1. The bone-on-bone compressive forces are shown as the muscle and gravitational force (ground reaction force) contributions to this load. As can be seen from Figure 1, extensor muscle force is by far the greatest contributor to the joint compressive force during level (70%) and downhill (80%) walking.

From the estimated knee joint loads, we calculated the stress on the tibial plateau for a 70 kg female in several walking tasks, plotted against varying tibiofemoral contact area, assuming a uniform pressure distribution. The results are shown in Figure 2, together with an indication of the yield range for ultra high molecular weight polyethylene (UHMWPE). It is clear that in order to obtain stress levels which are safely below the yield point of UHMWPE for all

Fig. 1

Mean tibiofemoral joint loadings of the six female subjects during the stance phase in level and downhill walking. The values are reported in multiples of body-weight (BW) and normalised to 100% of the stance time. Heel strike occurs at 0% and toe-off at 100%. The total uncertainty (SEOM) is indicated for the peak tibiofemoral compressive force values.

Fig. 2

Plot of the tibial plateau stress versus the contact area for different daily activities for a 70 kg female subject. A uniform stress distribution is assumed. The range of tibiofemoral contact area of current knee prostheses³¹ (TK), a knee joint after meniscectomy³⁰ (MK) and a natural knee joint³⁰ (NK) are shown. The horizontal bar indicates the yield range of polyethylene.

daily activities a contact area greater than 400 mm² is required.

DISCUSSION

This study demonstrates that the muscle forces contribute 80% of the maximum bone-on-bone force during downhill walking and 70% of the maximum bone-on-bone force during level walking. The magnitude of the ground reaction force is not the best predictor of the joint load, the muscle moments are more reliable. These patellar ligament forces are based on calculations of the net muscle moment of force acting about the knee, which is a limitation of the inverse dynamics approach. Quadriceps effort required to overcome antagonistic effects of the knee flexor muscles is not included. Electromyographic activity, recorded from our subjects during their downhill walking, clearly indicated the presence of hamstring, gastrocnemius and quadriceps muscle co-activity during the stance phase.¹⁹ The measures of joint compression reported here are therefore conservative estimates but still exceed eight times body-weight for downhill walking.

Force values equivalent to three to four times BW have previously been used in most biomechanical tests evaluat-

ing total knee replacements.^{11,20,21} Estimates of the tibiofemoral bone-on-bone forces in our study were close to four times BW even during level walking and more than eight times BW during downhill walking. Collins⁹ calculated the knee joint loads during level walking using an optimisation method which incorporated muscle coactivation of agonists and antagonists. He concluded that the tibiofemoral loads range from 3.9 to 6.0 times BW. Jefferson et al¹⁰ found that the maximum tibiofemoral loads are up to 6.3 times BW, while Wyss et al²² report values ranging from 2.5 to 5 times BW. Our results for level walking are well within the limits of these predictions. As the experimental set-up for level and downhill walking did not change, the calculated loads for downhill walking, allowing for the angulation of the plate, present a valid comparison.

Loads for biomechanical evaluation of patellar components have been considered to be in the range of 0.7 to 2 times BW.²³ Some researchers, in order to evaluate different designs, assumed loads as low as 0.15 times BW for level walking and 2.2 times BW for walking downstairs.²⁴ Recent research indicates higher loadings; for level walking patellofemoral joint forces of 1.3 to 1.8 times BW,^{16,25} for downstairs walking 5.5 times BW¹⁶ and for downhill walking 5 to 7 times BW have been suggested.²⁵ For some sports activities such as jumping²⁶ or weightlifting²⁷ the loads imposed on the patellofemoral joint are close to 20 times BW.

A review of the available literature indicates that the majority of authors use the lowest tibiofemoral and patellofemoral joint loads reported for the evaluation of contact stresses in total joint replacement.^{5,11,21,23,24,28} In order to improve the design of total knee replacements, it is necessary to adopt higher tibiofemoral and patellofemoral loads.

Contact area has a very profound effect on joint stress (Fig. 2). The reported average contact area of a natural knee joint ranges from 765 mm² to 1150 mm².³⁰ After complete medial and lateral meniscectomy the tibiofemoral contact area is approximately 520 mm², depending on the load.^{29,30} Assuming a uniform load distribution and a load of eight times BW the estimated stress on the articular cartilage is only about 10 MPa for a knee joint without menisci and less than 5 MPa for a healthy knee. The contact area of most total knee prostheses is between 80 and 300 mm² depending on the load, flexion angle and design,³¹ leading to contact stresses on the UHMWPE inlay as high as 60 MPa; this exceeds the yield point of 20 MPa for UHMWPE.

A contact area of approximately 400 mm² is necessary to avoid stresses to the polyethylene inlay that are above the yield point of 20 MPa. This contact area should be maintained throughout a flexion range of 0° to 60° to accommodate the high loads of downhill and downstairs walking. Congruent prostheses significantly reduce polyethylene wear, and line or point contact should be avoided.^{32,33}

In summary, the design of knee replacements should allow for the much higher joint loadings now being estimated through gait analysis if severe wear is to be avoided.

Table I. Reported **tibiofemoral** joint loads for several daily activities

Author(s)	Activity	Tibiofemoral joint load (body-weight multiples)
Ericson and Nisell ³⁴	Cycling	1.2
Morrison ⁸	Level walking	3.0
Harrington ³⁵		3.5
Present study		3.9
Collins ⁹	Downstairs walking	3.9 to 6.0
Morrison ⁷		3.8
Andriacchi ³⁶		6
Morrison ⁷	Downhill walking	4.5
Present study		8
Ellis ³⁷	Rising from chair	3.2
Nisell ¹⁶	Isokinetic knee extension at 30°/sec	9
Dahlkvist ³⁸	Squat descent	5.6
Collins ³⁹	Weightlifting (120 kg)	up to 24

Table II. Reported **patellofemoral** joint loads for several daily activities (NB: where the data for the joint load calculations were taken from secondary sources then these are indicated in parentheses)

Author	Activity	Patellofemoral joint load (body-weight multiples)
Ericson et al ⁴⁰	Cycling	1.2
Reilly and Martens ⁴¹	Level walking	0.5
Matthews ⁴² (Morrison ⁸)		0.7
Nisell ¹⁶ (Boccardi ⁴³)		1.3
Kuster ²⁵	Stair ascending	1.8
Nisell ¹⁶ (Andriacchi ³⁶)		2.1
Reilly and Martens ⁴¹		3.3
Matthews ⁴² (Morrison ⁴⁴)		2.5
Nisell ¹⁶ (Andriacchi ³⁶)	Stair descending	5.6
Reilly and Martens ⁴¹		3.3
Matthews ⁴² (Morrison ⁴⁴)		2.5
Matthews ⁴² (Morrison ⁴⁴)	Downhill walking	1.8
Kuster ²⁵		7.0
Ellis ³⁷	Rising from chair	3.1
Nisell ¹⁶ (Winter ⁴⁵)	Jogging	7.0
Dahlkvist ³⁸	Squat descent	7.6
Huberti ⁴⁶	Isometric contraction at 90° flexion	6.5
Nisell ⁴⁷	Isokinetic exercise at 30°/sec	12.0
Nisell ¹⁶ (Wahrenberg ⁴⁸)	Kicking	7.6
Zernicke ²⁷	Patellar tendon rupture	17.5 - 25

No benefits in any form have been received or will be received from a commercial party related directly or indirectly to the subject of this article.

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