The relationship of the angle of immobilisation of the knee to the force applied to the extensor mechanism when partially weight-bearing

A GAIT-ANALYSIS STUDY IN NORMAL VOLUNTEERS

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We describe the influence of the angle of immobilisation during partial weight-bearing on the forces across the extensor mechanism of the knee. Gait analysis was performed on eight healthy male subjects with the right knee in an orthotic brace locked at 0°, 10°, 20° and 30°, with the brace unlocked and also without a brace. The ground reaction force, the angle of the knee and the net external flexion movement about the knee were measured and the extensor mechanism force was calculated.

The results showed a direct non-linear relationship between the angle of knee flexion and the extensor mechanism force. When a brace was applied, the lowest forces occurred when the brace was locked at 0°. At 30° the forces approached the failure strength of some fixation devices. We recommend that for potentially unstable injuries of the extensor mechanism, when mobilising with partial weight-bearing, the knee should be flexed at no more than 10°.

Injuries to the extensor mechanism include fractures of the patella and rupture of the quadriceps and patellar ligament. They account for between 1% and 2% of all musculoskeletal injuries. Treatment of these conditions makes considerable demands on operating time, rehabilitation and follow-up resources. If these injuries are not managed appropriately, complications such as stiffness and osteoarthritis of the knee can occur.

Irrespective of whether such injuries are treated surgically or not, they are often managed by partial weight-bearing and the use of a splint for approximately six weeks. For most cases it is imperative that, in the early stages after injury, the tensile forces acting across the extensor mechanism are kept to a minimum. In some operatively-managed cases, such as with tension-band wiring of patellar fractures, it is desirable that some force acts through the extensor mechanism.

We are not aware of any study of gait-analysis which has related the angle of immobilisation of the knee to the forces acting on the extensor mechanism during restricted weight-bearing. Our aim was to identify the angle of immobilisation in partially weight-bearing normal subjects which resulted in the minimum force across the extensor mechanism. This information can then be used in the management of partially weight-bearing patients with injury to the extensor mechanism to produce the biomechanically optimum treatment.

Patients and Methods
After approval from the local ethics committee eight healthy male volunteers with a mean age of 28.6 years (25 to 37) and mean height and weight of 1.74 m (1.60 to 1.85) and 75.6 kg (64 to 86) respectively, were recruited to the study and provided informed consent. They had no previous history of musculoskeletal or neurological symptoms in the lower limb and were not participating in any regular programme of exercise.

Immobilisation. Five different types of knee immobilisation were evaluated using a MediROM orthotic knee brace (Medi UK Ltd, Hereford, United Kingdom) unlocked and locked in 0°, 10°, 20° and 30° of knee flexion. The knee was also studied without any brace.


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Collection of data. Information regarding force and three-dimensional movement was collected using a single Kistler force plate (Kistler, Alton, United Kingdom) sampling at 200 Hz and an eight-camera Qualisys ProReflex movement analysis system (Qualisys Medical AB, Gothenburg, Sweden) sampling at 100 Hz. Retroflective markers were placed at the anterior and posterior superior iliac spine, the centre of the greater trochanter, the lateral and medial joint lines, the lateral and medial malleolus, the calcaneum and the base of the fifth, first and second metatarsals to determine the position and orientation of each segment in accordance with the Calibrated anatomical system technique (CAST) protocol. The movements of the pelvis, thigh and shin were recorded by applying cluster plates at these sites which were positioned so that there was no interference with the application of the brace. Three tests were carried out for each type of immobilisation. In order to improve repeatability, only one observer (WSK) took the measurements using the same brace and laboratory set-up for all the data collection. In addition, the right leg was studied in all subjects. The trials were carried out in a random order to improve internal reliability. The subjects walked on the force plate at a self-selected speed. The cluster plates were left in situ between the application and removal of the brace to minimise errors resulting from malpositioning of the markers.

Data, including the subjects’ height and mass, were exported to Visual3D movement analysis software (C-motion Inc, Rockville, Maryland) in which a Butterworth fourth-order filter smoothed the data with a cut-off frequency of 6 Hz for data on movement and 15 Hz for those on force. The mean knee flexion angles and net external flexion movements about the knee were calculated for the stance phase using this software. The flexion movements were calculated using inverse dynamics. During the measurements, the subjects were asked to put half their weight through the weight-bearing leg and half through elbow crutches. The weight going through the leg was monitored using the Kistler force plate, and the remaining weight was assumed to pass through the elbow crutches.

Calculations. The three principal coplanar forces acting on the tibiofemoral joint are the tibiofemoral joint reaction force (J), the extensor mechanism force (E), which acts through the patellar ligament, and the ground reaction force (W) as shown in Figure 1. The sum of all movements about the knee at the tibiofemoral joint in a static situation is zero. Therefore, Wa = Eb, where a and b are the movement arms for the forces W and E, respectively.

The extensor mechanism force (E) was calculated for all six types of immobilisation by dividing the flexion movement by the length of the extensor mechanism movement arm (b). The values for this length were calculated using the formula:

\[ b = 0.0367x + 3, \]

where \( x \) is the knee flexion angle in degrees. Statistical analysis. Two-tailed unpaired and paired \( t \)-tests were performed on the total body-weight and the knee flexion angles. A one-way analysis of variance (ANOVA) was carried out to compare the within-subject variance between each type of immobilisation. The level of statistical significance applied was \( p \leq 0.05 \), after Bonferroni’s correction.

Results

The normalised vertical component of the ground reaction force expressed as a proportion of the total body-weight is presented in Table I. There was no statistically significant difference between the actual proportion of the total body-weight going through the leg for any type of immobilisation or when the brace was not used (ANOVA, \( p = 0.61 \)).

The knee flexion angle in the sagittal plane was measured from the kinematic measurements and the mean result was compared with the angle set on the brace (Table I). Significant differences between the set and the mean of the measured flexion angles were seen when the brace was set at 0˚ (\( t \)-test, \( p = 0.003 \)) and 10˚ (\( t \)-test, \( p = 0.030 \)). There was also a significant difference between the mean knee flexion angle...
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The flexion angles obtained from the kinematic data, the calculated lengths of the extensor mechanism movement arm, the net external flexion movement in the sagittal plane normalised to the mass of the subject and the calculated extensor mechanism force for each of the six types of immobilisation are shown in Table I. A test of within-subjects effects showed a statistically significant difference between the extensor mechanism forces for the six types of immobilisation (ANOVA, p < 0.001). The results of pair-wise comparisons are given in Table II.

Discussion

The quantitative kinematic analysis used in our study has previously been validated as being a useful instrument for assessing joint function during movement. It can be used in combination with force-plate data to determine the forces causing the movement.

The clinical practice of instructing patients to bear half of their total body-weight subjectively is widespread and was used in our study. Our subjects were able to put half their body-weight through the test side without difficulty, although we accept that they were not experiencing any pain which might have been associated with a recent injury. The knee flexion angles measured using the kinematic data showed a similar trend to that set at the brace, but they did not always correspond. The discrepancy was greatest when the brace was set at 0° and least at 30° of flexion. It is unlikely that the kinematic results at low angles were inaccurate since the equipment used in our study had previously been well validated, and at higher angles the measurements were similar to those set at the brace. The kinematic data also showed that the unlocked brace did not allow the wearer to obtain the same low knee flexion angles that are recorded when there was no brace. These findings suggest that there is scope for improvement in the design and calibration of knee braces to allow more reliable control of the knee at low angles of flexion.

The values of the length of the extensor mechanism movement arm derived from a linear equation involving the knee flexion angle. In calculating this length we relied on the extrapolation of published data, which have previously been used to calculate the extensor mechanism movement arm. However, the exact values are likely to

Table I. The mean (range) kinematic and kinetic measurements and calculations for all six types of immobilisation

<table>
<thead>
<tr>
<th>Amount of flexion set at the orthotic knee brace</th>
<th>0°</th>
<th>10°</th>
<th>20°</th>
<th>30°</th>
<th>Unlocked</th>
<th>No brace</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean ground reaction force as a proportion of the total body-weight (range)</td>
<td>0.56 (0.36 to 0.78)</td>
<td>0.56 (0.34 to 0.73)</td>
<td>0.54 (0.32 to 0.75)</td>
<td>0.54 (0.31 to 0.78)</td>
<td>0.52 (0.31 to 0.91)</td>
<td>0.56 (0.34 to 0.82)</td>
</tr>
<tr>
<td>Mean of knee flexion measured using kinematic data (°) (range)</td>
<td>9.8 (17.4)</td>
<td>16.3 (7.4 to 23.4)</td>
<td>22.5 (16.2 to 31.3)</td>
<td>28.7 (25.1 to 36.3)</td>
<td>13.6 (5.3 to 21.4)</td>
<td>5.0 (2.0 to 17.8)</td>
</tr>
<tr>
<td>Mean extensor mechanism movement arm (cm) (range)</td>
<td>3.36 (3.27 to 3.86)</td>
<td>3.60 (3.27 to 3.86)</td>
<td>3.83 (3.60 to 4.15)</td>
<td>4.05 (3.92 to 4.33)</td>
<td>3.50 (3.19 to 3.79)</td>
<td>3.18 (2.93 to 3.65)</td>
</tr>
<tr>
<td>Mean normalised net external flexion movement about the knee in the sagittal plane (Nm/kg) (range)</td>
<td>0.06 (0.00 to 0.38)</td>
<td>0.12 (0.00 to 0.38)</td>
<td>0.23 (0.00 to 0.38)</td>
<td>0.27 (0.00 to 0.38)</td>
<td>0.11 (0.00 to 0.38)</td>
<td>-0.01 (0.00 to 0.38)</td>
</tr>
<tr>
<td>Mean normalised extensor mechanism force (N/kg) (range)</td>
<td>1.66 (4.87)</td>
<td>4.48 (9.95)</td>
<td>5.85 (9.95)</td>
<td>6.58 (9.95)</td>
<td>3.02 (6.81)</td>
<td>-0.41 (8.18)</td>
</tr>
</tbody>
</table>

Table II. The statistical significance* (p-value) for paired comparisons of normalised extensor mechanism forces for the six types of immobilisation

<table>
<thead>
<tr>
<th>Immobilisation</th>
<th>0°</th>
<th>10°</th>
<th>20°</th>
<th>30°</th>
<th>Unlocked</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>0.070</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10°</td>
<td>0.026</td>
<td>1.000</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20°</td>
<td>0.003</td>
<td>0.196</td>
<td>1.000</td>
<td></td>
<td></td>
</tr>
<tr>
<td>30°</td>
<td>1.000</td>
<td>0.317</td>
<td>0.461</td>
<td>0.073</td>
<td></td>
</tr>
<tr>
<td>Unlocked</td>
<td>1.000</td>
<td>0.607</td>
<td>0.111</td>
<td>0.107</td>
<td>1.000</td>
</tr>
<tr>
<td>No brace</td>
<td>1.000</td>
<td>0.607</td>
<td>0.111</td>
<td>0.107</td>
<td>1.000</td>
</tr>
</tbody>
</table>

* analysis of variance paired comparisons

Graph showing the relationship between the degree of knee flexion measured using kinematic data and the normalised extensor mechanism force. The mean value and SEM are shown as black diamonds and error bars, respectively. The line of best-fit shows that the relationship is not linear.
vary from individual to individual and the use of documented anthropometric data should be exercised with caution.

We found that the extensor mechanism force increased with increasing knee flexion. The negative values in early flexion were a result of the knee flexion movement passing in front of the centre of movement of the tibiofemoral joint in the sagittal plane and, in effect, causing a knee extension movement. Figure 2 shows that there is a direct relationship between the knee flexion angle and the extensor mechanism force. The relationship is not linear and the gradient of increase lessens with increasing flexion. Importantly, we found that lower flexion angles corresponded to lower normalised extensor mechanism forces.

Our study had some limitations. We used a simplified model of the knee which ignored co-contractions from muscles including the hamstrings and the forces as a result of the structure of the brace itself. The measured net external flexion movement was thus not a free muscle movement but, for simplification, was considered to be so in our calculations. The determination of anatomical landmarks is always subject to some error and this is a known limitation of our method. However, error from movement of skin was not a consideration in our study since we avoided the use of skin markers and used cluster plates. The latter were placed at predetermined locations which were known not to interfere with the application of the knee brace.

The force required to disrupt any form of operative or conservative treatment in injury to the extensor mechanism has been poorly researched. It will vary among patients, pathologies and methods of treatment. The use of suture anchors for the repair of ligaments or tendons and techniques of fixation of patellar fractures give tensile pull-out strengths of around 600 N.\textsuperscript{14-16} The extensor mechanism force measured for immobilisation at 0° in our study for a 70 kg patient is comfortably below this value. For the same patient, immobilisation at 30° may produce extensor mechanism forces which approach this value when partially weight-bearing.

Our study has raised a number of issues concerning the rehabilitation of patients with injury to the extensor mechanism. For most of such injuries we recommend the use of a brace to prevent excessive tensile forces resulting from excessive knee flexion and associated extensor mechanism forces. The use of an unlocked brace and no brace was nevertheless included in the study to provide a baseline. For repaired injuries to tendons or ligaments we recommend maximum immobilisation of 10° if the patient is to be allowed to bear weight partially. This may be increased in injuries in which tensile forces are desirable. Our study has not been able to give guidance on the use of free movement within a restricted range or in the non-weight-bearing situation. We therefore recommend that further investigation is necessary in these circumstances.

Clinicians treating injury to the extensor mechanism should carefully consider how their splintage devices are applied, set and used in each individual case.

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References