Biomechanical analysis of the walking pattern in two different types of ACL patients

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Introduction
Differences between the walking patterns of anterior cruciate ligament deficient (ACLD) patients and healthy subjects have been reported. However, various walking patterns in ACLD patients have been observed. Thus, the walking pattern of ACLD patients is still open to discussion. The reason for this may be that different types of ACLD patients walk according to different patterns (2). Rudolph et al 1998 observed differences between the walking patterns of ACLD patients who were able to maintain a high activity level (copers) despite their injury and those who were unable to return to the same activity level as before the injury (non-copers). However, they did no include a control group in their study and it is therefore unclear if and how the walking patterns of the copers and non-copers differ from a normal gait pattern. The purpose of the present study was to evaluate the walking pattern of the ankle, knee and hip joint in healthy subjects and in two types of ACLD patients, copers and non-copers, using an inverse dynamics approach and electromyography. Furthermore, the walking patterns observed in the injured and uninjured leg of the ACLD patients were compared to determine whether the ACL injury caused asymmetry.

Methods
Nineteen males with complete unilateral ACL deficiency participated in the study. All of them had been through rehabilitation programs for at least 6 months after their injury. The patients were separated into two groups consisting of those who experienced a good knee function (copers, n=9) and those who experienced a poor knee function after injury (non-copers, n=10). The copers (weight: 76.7±14.3 kg, height: 1.81±0.06 m, age: 28.3±6.1 years, mean±SD) were able to return to their normal activity level despite their injury while the non-copers (weight: 80.4±6.7 kg, height: 1.79±0.05 m, age: 31.7±5.9 years, mean±SD) were unable to return to the same activity level as before their injury. Nineteen healthy male subjects (weight: 76.7±6.6 kg, height: 1.82±0.04 m, age: 31.0±4.5 years, mean±SD) were selected as controls for the gait analysis. Electromyography (EMG) was recorded for 10 of the control subjects (weight: 77.5±7.9 kg, height: 1.82±0.05 m, age: 31.0±2.8 years, mean±SD). There were no statistically significant differences between the groups with regard to weight, height and age.

The subjects were asked to walk across two force platforms (AMTI, OR6-5-1) at a speed of 4.5 km/h. The speed was controlled by photocells, which made it possible to teach the subject to approach 4.5 km/h.

Fifteen small reflecting spherical markers (12-mm diameter) were placed on the subjects according to the marker set-up Vaughan (3). Five video cameras (Panasonic WV-GL350) operating at 50 Hz were used to record the movements. The video signals and the force plate signals were synchronized electronically with a custom built device. The device put a visual marker on one video field from all cameras and at the same time triggered the analogue-to-digital converter, which sampled the force plate signals at 1000 Hz. The subjects triggered the data sampling and synchronization when they passed the first photocell. The video sequences were digitized and stored on a PC. Three-dimensional co-ordinates were reconstructed by direct linear transformation using the Ariel Performance Analysis System (APAS). Prior to the calculations, the position data were digitally low-pass filtered by a fourth order Butterworth filter with a cut-off frequency of 6 Hz, and the 1000 Hz force plate signals were reduced to fit the 50 Hz video signals.

The angular position of the ankle, knee and hip joint was calculated to describe the movements in the sagittal plane. Zero degrees defined the anatomical position and positive values reflected hyperextension of the hip and knee joints and ankle plantar flexion. Internal flexor and extensor joint moments about the ankle, knee and hip were calculated using a three-dimensional inverse dynamics approach (3). As a measure of the total extensor moment developed during the stance phase, the support moment was calculated by a summation of the ankle, knee and hip joint moments (4). The joint power was calculated by multiplying the joint moments and the joint angular velocity. Positive and negative work were then estimated by calculating the area under the power curves. These phases correspond to phases A1 and A2.
for the ankle, K1 and K2 for the knee and H1 for the hip as described by Winter (4). The peak values of
the joint moment and power curves as well as negative and positive work developed in the first half of the
stance phase were used as input parameters for the statistical analysis. The average position of the joint
angles and peak values of the knee flexion in the stance phase were also tested statistically. MATLAB
was used for all calculations.

Data obtained from the right leg of the control group and the injured (ACLD) and uninjured (UNI) leg
of the two patient groups were analyzed. Six gait cycles were normalized and averaged for each subject.
Only the stance phase was analyzed. The joint moments, power and work were normalized to body mass.
Ensemble averages were then calculated for the non-copers, copers and the control group using the mean
value for each individual subject.

EMG was obtained from quadriceps (m. rectus femoris, m. vastus lateralis, and m. vastus medialis)
and the medial and lateral part of the hamstring muscle on right leg of the controls and the injured leg of
the ACLD subjects. The EMG signals were digitally high- and low-pass filtered (Butterworth fourth order
zero-lag digital filter, cut-off frequencies 20 Hz and 500 Hz, respectively), full-wave rectified and low-pass
filtered at 15 Hz to perform linear envelopes. Linear envelopes from 20 walking trials were used to
calculate the average EMG of each muscle for each subject. All signals were expressed in microvolt and
normalized in 12 slices representing the stance phase. The EMG activity of quadriceps (represented by
RF, VL, VM) and hamstring muscles (represented by the medial and lateral part of the hamstrings) was
calculated by adding the individual EMG signals of the muscles in each muscle group. Furthermore,
the total EMG activity of the hamstrings and quadriceps was calculated by adding the two signals.

The t-test for paired samples was used for statistical comparisons between the ACLD and UNI legs of
the non-copers and copers. Statistical comparisons between the non-copers (NON), copers (COP) and
control group (CON) were obtained by using a one-way analysis of variance test (ANOVA). In cases with
significant differences, the Student-Newman-Keuls method was used to locate the differences. The level
of significance was set at 5%. All results are presented as means±SD.

Results & Discussion

The average walking velocity for all the subjects was 4.5±0.2 km/h and there was no difference in the
walking velocity between the three groups (Table 1). The three groups walked with identical step length,
cadence and relative swing and stance time (Table 1).

<table>
<thead>
<tr>
<th>Variable</th>
<th>Non-copers (n=10)</th>
<th>Copers (n=9)</th>
<th>Control (n=19)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Velocity (km/h)</td>
<td>4.5 (0.1)</td>
<td>4.6 (0.2)</td>
<td>4.5 (0.2)</td>
</tr>
<tr>
<td>Step length (m)</td>
<td>0.75 (0.02)</td>
<td>0.76 (0.05)</td>
<td>0.74 (0.02)</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>101 (2)</td>
<td>102 (3)</td>
<td>103 (4)</td>
</tr>
<tr>
<td>Swing time (%)</td>
<td>37.9 (1.6)</td>
<td>36.8 (1.5)</td>
<td>36.6 (1.5)</td>
</tr>
<tr>
<td>Stance time (%)</td>
<td>62.1 (1.6)</td>
<td>63.2 (1.5)</td>
<td>63.4 (1.5)</td>
</tr>
</tbody>
</table>

There were no differences between the walking pattern of the injured and uninjured leg in the ACLD
patients. However, the analysis of the walking kinematics and dynamics revealed that the copers and non-
copers used different walking patterns and that these differed from the walking pattern observed in the
control subjects (Fig 1). The copers walked with more flexed knee joints in the first half of the stance
phase than the non-copers and the controls. The non-copers walked with joint kinematics similar to the
control subjects. Statistically significant differences were observed in the magnitude of the knee and hip
joint moments. The non-copers walked with a reduced peak knee joint moment in the first half of the
stance phase compared to the controls. There was no difference in the peak knee joint moment between
the copers and the controls. The peak hip joint moment was larger in the copers than in both the non-
copers and controls. The positive work developed by the hip extensors was higher in the copers and non-
copers than in the control subjects. The contribution of the knee and hip joint moment to the support
moment was larger in the copers than in the non-copers and the controls, which resulted in a larger peak
support moment in the copers.
The amplitude of the quadriceps was higher at 30% of the stance phase in the copers (88.2±58.6 µV) than in the non-copers (49.8±23.5 µV) and controls (44.6±6 µV). No significant difference was observed in the amplitude of the hamstring muscles between the groups. However, the total sum of the hamstrings and quadriceps EMG was higher in the copers than in the non-copers at 25% (240.0±151.8 (COP), 144.0±46.2 (NON) µV), and 30% (112.9±67.5 (COP), 68.2±30.4 (NON) µV) of the stance phase.

It is possible that the greater knee flexion in the copers created better conditions for co-contraction of the quadriceps and hamstring muscles. Computer simulations of the knee have shown that co-contraction of the quadriceps and hamstring muscles loads the ACL from full extension to 22° of flexion while simultaneous contractions of the muscles can unload the ACL at knee flexion angles above 22° (1). The peak knee flexion angle was 27.1° in the copers and 21.6° in the non-copers, which suggests that the copers were able to stabilize their knee joint by co-contraction. The EMG results showed that the collective hamstrings and quadriceps EMG activity was higher in the copers than in the non-copers and the control subjects, which supports the fact that co-contraction was increased in the copers. Co-contraction could not reduce the anterior shear of the tibia in the non-copers because of the unaltered knee joint kinematics. In addition, the EMG results indicated that the non-copers walked with the same amount of co-contraction as the controls. Thus, the non-copers had to reduce the knee extensor moment to reduce the stress on the knee joint. It is possible that the non-copers would achieve a better knee function if they learned to walk as the copers. Further investigation is needed to evaluate this question.

References