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Downhill walking: a stressful task for the anterior cruciate ligament?

A biomechanical study with clinical implications

M. Kuster[1], G. A. Wood[2], S. Sakurai[3], G. Blatter[1]

[1] Klinik für Orthopädische Chirurgie, Kantonsspital, St. Gallen, Switzerland
[2] Department of Human Movement, The University of Western Australia, Perth, Australia
[3] Research Centre of Health, Fitness and Sports, Nagoya University, Nagoya, Japan

Abstract. Accelerated rehabilitation after anterior cruciate ligament (ACL) reconstruction has become increasingly popular. Methods employed include immediate extension of the knee and immediate full weight bearing despite the risks presented by a graft pull-out fixation strength of 200–500 N. The purpose of this study was to calculate the tibiofemoral shear forces and the dynamic stabilising factors at the knee joint for the reasonably demanding task of downhill walking, in order to determine whether or not this task presented a postoperative risk to the patient. Kinematic and kinetic data were collected on six male and six female healthy subjects during downhill walking on a ramp with a 19% gradient. Planar net joint moments and mechanical power at the knee joint were calculated for the sagittal view using a force platform and videographic records together with standard inverse dynamics procedures. A two-dimensional knee joint model was then utilised to calculate the tibiofemoral shear and compressive forces, based on the predictions of joint reaction force and net moment at the knee. Linear envelopes of the electromyographic (EMG) activity recorded from the rectus femoris, gastrocnemius and biceps femoris muscles were also obtained. The maximum tibiofemoral shear force occurred at 20% of stance phase and was, on average, 1.2 times body weight (BW) for male subjects and 1.7 times BW for female subjects. The tibiofemoral compressive force was 7 times BW for males and 8.5 times BW for females during downhill walking. The hamstring muscle showed almost continuous activity throughout the whole of the stance phase. The gastrocnemius muscle had its main activity at heelstrike, with a second brust during the late stance phase. Knee joint shear force predictions of approximately 1000 N for a 70-kg subject greatly exceed the strength of a typical ACL graft fixation and muscular stabilisation of the knee is therefore vital to joint integrity. The hamstring muscle shows almost continuous activity during the stance phase and thereby affords some stability, but the gastrocnemius is also seen to be an important stabiliser of the knee joint in the presence of increased shear forces during early stance. Associated stability to the knee joint is indicated by compressive loadings of 7–8 times BW across the tibiofemoral joint. Whereas under normal circumstances there is sufficient dynamic joint stabilisation during downhill walking, the muscular impairment often arising postoperatively from disturbed proprioception could endanger an ACL graft. Therefore downhill walking should be avoided during the postoperative phase in order to protect the reconstruction.

Key words: Anterior cruciate ligament – Downhill walking – Gait analysis – Electromyography – Rehabilitation

Introduction

Accelerated rehabilitation after anterior cruciate ligament (ACL) reconstruction has become increasingly popular [18]. Methods employed include immediate extension of the knee joint and immediate full weight bearing. However, our recent research has shown that downhill walking is a demanding task for the knee joint in terms of joint moments and muscle power [9]. Early work by Morrison [13, 14] considered the forces in the knee joint during downhill walking amongst other activities. His results for downhill walking, however, are based on an investigation of three healthy subjects only without controlled cadence. Since slope walking plays an important role in our daily activities, further biomechanical data on the forces acting at the tibiofemoral joint as well as information on dynamic factors which stabilise the knee joint during increased stress were considered important. This paper is concerned with the shear forces and the muscular stabilising factors at the knee joint during downhill walking and

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Correspondence to: M. Kuster MD, Klinik für Orthopädische Chirurgie, Kantonsspital, CH-9007 St. Gallen, Switzerland

level walking in order to determine whether or not this tasks present a postoperative risk to the graft after ACL reconstruction.

Subjects and methods

Subjects

Twelve healthy young adults (six male and six female) were selected for this study. The subjects were chosen on the basis of having no lower limb or back pathology nor any apparent leg length inequality. They ranged in age from 23 to 37 years (mean 27.9 years), in height from 158 to 187 cm (mean 171.2 cm) and in weight from 49 to 90 kg (mean 70.8 kg). Prior to participation all subjects signed a consent form in accordance with procedures required by the University of Western Australia for experimentation involving human subjects.

Experimental setup

A standardised setup for a reliable comparison of downhill and level walking was designed for the purposes of this study. For downhill walking a dismountable ramp of 6 m length with a grade of 19% was specially constructed. Located at a point 2 m up this ramp was an aluminium plate with surface dimensions the same as a Kistler model 9281B force platform. This plate was independently supported upon a rigid aluminium scafffold that bolted to the four corners of the Kistler force platform located directly below. Ground reaction force and centre of pressure measures could thereby be obtained during downhill walking in a manner not dissimilar to that for level walking [17]. Three-dimensional (3D) kinematic data of each subject's gait were obtained using a two-camera 60-Hz video-based motion analysis system (APAS, Ariel Dynamics, Inc.). One camera was set at 90° to the sagittal plane and the other at 60° to the first in order to obtain a dorsal perspective. Heel strike was identified on the video record by a stroboscope.

Reflective markers were placed over the location of each joint of the lower extremity. The control of gait cadence which is known to be necessary in order to derive meaningful group data [23] was achieved by the use of a metronome set to a frequency of 120 steps/min.

Electromyographic (EMG) data were collected from the rectus femoris, the biceps femoris and the gastrocnemius muscles using bipolar surface electrodes. The raw EMG signals were amplified, bandpass filtered (3 dB down at 3 Hz and 1 kHz) and sampled at 500 Hz on an IBM-PC compatible computer. A pressure switch inserted into the heel of the subject's shoe was used to synchronise EMG data with kinematic measures.

Each subject performed several practice trials of both downhill and level walking in order to become accustomed to the set step frequency and the laboratory situation. Data collection for one half of the subjects began with downhill walking trials, following which the ramp was immediately dismantled and level walking trials were then undertaken using the same markers, force platform and filming setup. The other half of the group were tested in reverse order. In this way proper comparisons between downhill and level walking were assured and test order effects were minimised. A minumum of five trials were performed by each subject on each task in order to obtain at least ten cycles of EMG data for ensemble average processing. The trial providing the best data in terms of correct cadence, natural walking form and good force platform results was used for kinetic and kinematic data processing.

Data processing

A standard cartesian coordinate system comprising a right-hand orthogonal triad fixed in the ground with the $+X_g$ axis forward and horizontal, the $+Y_g$ axis upward and the $+Z_g$ axis to the right and horizontal, as recommended by the International Society of Biomechanics (ISB) committee for Standardisation and Terminology, was used for kinematic and kinetic data processing. The centre of the force platform was taken as the origin of this system. The 'best' trial of each subject for downhill and level walking was digitised. The digitised marker position-time data were smoothed using a fourth-order Butterworth digital filter with a low-pass cut off of 7 Hz, and time derivates were then calculated from finite differences equations [22]. These kinematic data were then normalised to 100% of stride cycle and averaged across subjects.

Planar net joint moments at the knee joint were calculated for the sagittal view using force platform vertical and horizontal (F_y; F_x) and centre of pressure (CP) records based on standard inverse dynamics procedures and anthropometric values [22]. All kinetic data were normalised to 100% of stance phase from heelstrike to toe-off.

The EMG data were first high-pass filtered at 4 Hz to remove movement artefacts and then full-wave rectified. In order to overcome stride-to-stride variations in EMG patterns, ensemble averages of 10–19 strides were then computed for each subject. A representative group linear envelope of these averaged EMG records was then obtained by further averaging following normalisation of each individual ensemble average to 100% of the peak activity and 100% of stride time.

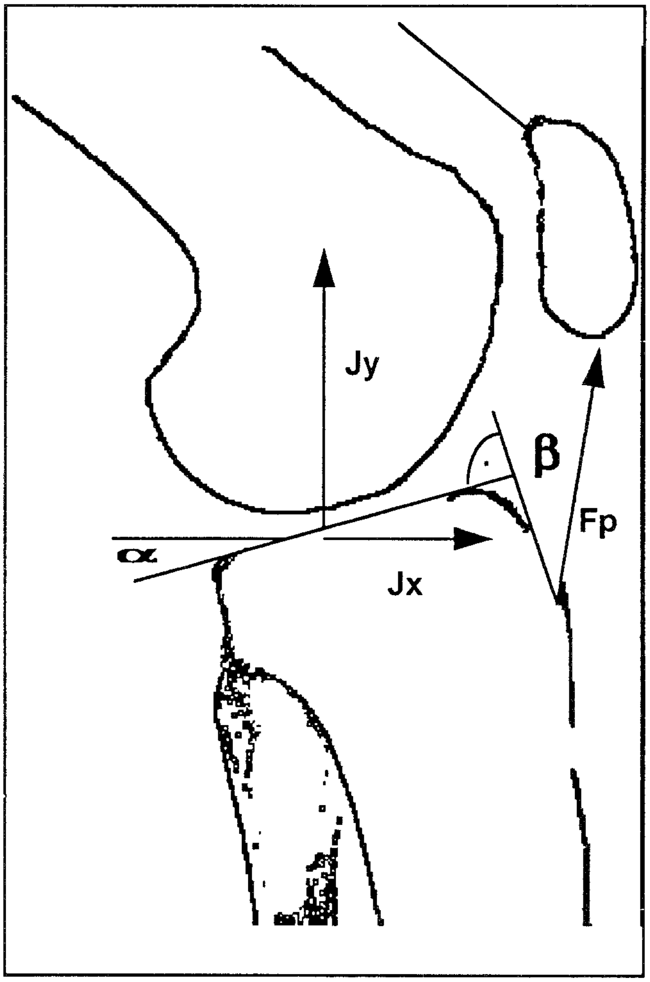

Fig. 1. Free body diagram of the knee joint. The tibiofemoral shear force (F_s) can be calculated from the joint reaction force J_x and J_y, the angle α of the tibial plateau to the horizontal plane, the patellar tendon force (F_p) and its angle β to the tibial plateau

Knee joint model (Fig. 1)

The tibiofemoral joint shear and compressive forces (F_s; F_c) were calculated from the joint reaction forces (J_x; J_y) and the patellar tendon force (F_p) occasioned by the net moments and inertia forces predicted to be acting about the knee. Knee joint musculoskeletal geometry was assumed to conform to the biomechanical data reported by Nisell [15]. However, in contrast to the approach of Nisell, the results reported here also taken into account the accelerations of body segments and the intersegmental effects between shank and foot.

Results

The EMG activity patterns of the quadriceps, hamstring and gastrocnemius muscles during the stance phase for downhill and level walking are shown in Fig. 2.

The *quadriceps muscle* showed for both downhill and level walking the main peak activity during the early stance phase.

Figure 2 shows the EMG activity during the stance phase only. Since the *hamstring muscle* had its peak activity during late swing phase for level and for downhill walking, this main activity is not presented in the figure. During downhill walking, however, distinct activity was seen at early and late stance phase with only a brief period of relative quiescence at mid stance, while for level walking only minimal or no activity occurred during the stance phase.

The *gastrocnemius muscle* had two main periods of peak activity during slope walking, with the first occurring at heelstrike and the second, smaller peak in late stance. During level walking this muscle was only minimally active at heelstrike and its main burst of activity occurred at late stance phase in preparation for push-off.

No significant differences were found between male and female subjects for any kinematic or kinetic measure derived from preliminary motion analysis. However, since Nisell's knee extensor moment arm values for females are significantly smaller than those of males [15], the calculated patellar tendon and therefore knee joint compression and shear forces were predictably larger for females.

The pattern of *shear force* (F_s) at the knee joint was predominantly anterior shear for downhill walking. For level walking an anterior shear force peak occurred at early stance phase followed by a posterior shear at late stance phase (Fig. 3). Peak values were less than body weight (BW) for level walking (7 N/kg for males and 9 N/kg for females), whereas they were almost 2 times higher during downhill walking (12 N/kg for males and 17 N/kg for females).

The tibiofemoral joint *compressive force* pattern for level walking had its main peak during early stance with a maximal force of 34 N/kg for male subjects and 40 N/kg for female subjects, whereas peak values for downhill walking were 71 N/kg for male subjects and 85 N/kg for female subjects.

Discussion

Morrison's work [13, 14] was the first and until now only study that has calculated joint forces and muscle power at the knee for downhill walking and level walking based on gait analysis. However, Morrison studied only three subjects and all were free to choose their own cadence; two of the three subjects reduced their step length as well as their cadence during downhill walking. Since it is now known that the knee joint moment is dependent on the speed of progression [1, 23], a valid comparison of downhill versus level walking is difficult from Morrison's data. Nevertheless, one of his subjects chose a cadence of 120 steps/min during both level and downhill walking and the muscle power and tibiofemoral joint compressive force values reported for that subject agree well with those of the present study in which there was adequate experimental control.

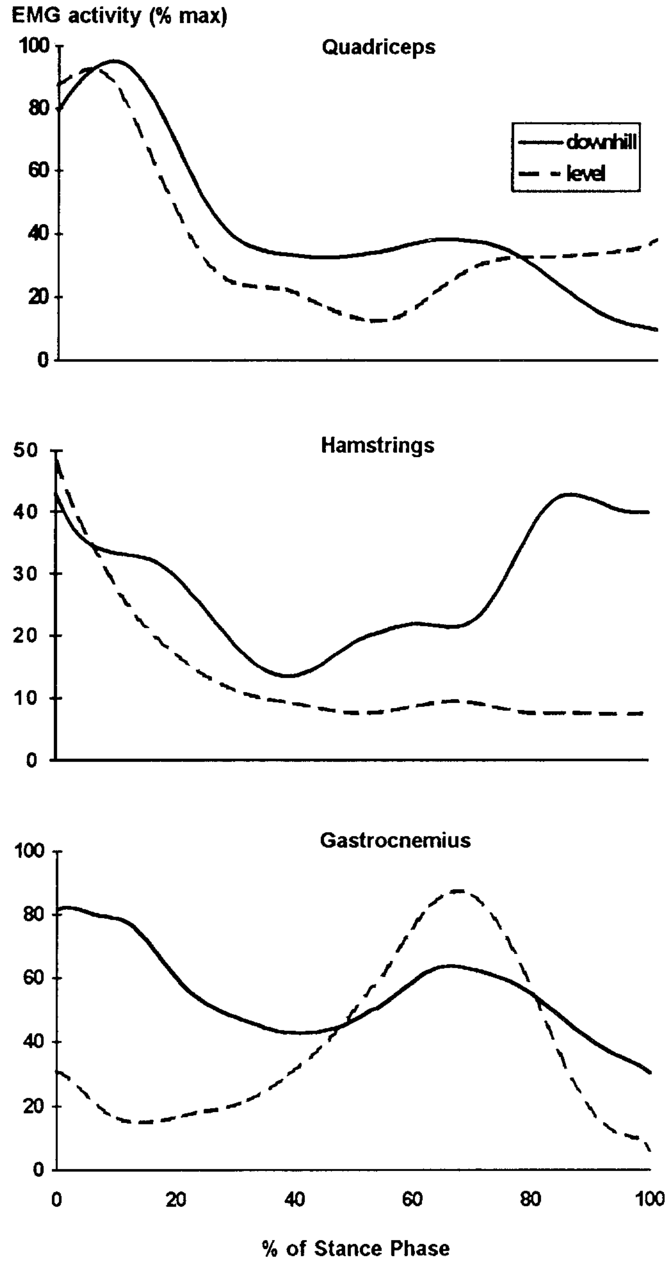

Fig. 2. Averaged EMG linear envelopes of the quadriceps, hamstrings and gastrocnemius muscles normalised to 100% of stance phase. Heelstrike occurs at 0% and toe-off at 100% of the stance phase. The level of EMG activity is normalised to subjects' peak activity during a cycle

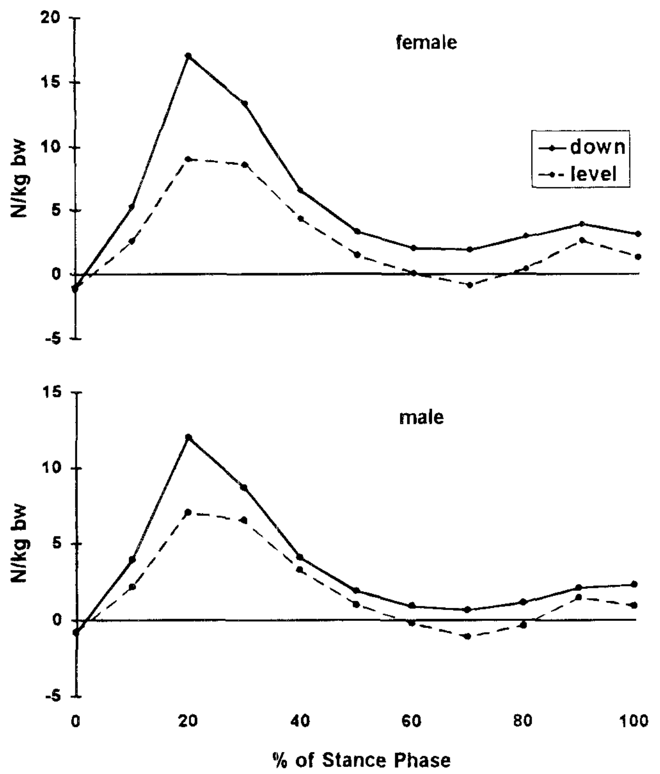

Fig. 3. The tibiofemoral shear force during downhill walking compared to level walking for male and female subjects. The values are reported in N/kg body weight and normalised to 100% of the stance phase. Heelstrike occurs at 0% and toe-off at 100%. A positive value indicates an anterior drawer force and a negative value a posterior drawer force of the tibia

It must be remembered that the moment of force calculated by the inverse dynamic procedure produces a net moment only. This means that any flexor moment produced by the hamstrings or gastrocnemius muscles must, in reality, be compensated for by a higher extensor moment of the quadriceps muscle, but this increased muscular effort is not reflected in the moment calculation.

Whereas for level walking the EMG activity of the hamstrings and gastrocnemius during the early stance phase is extremely small, this activity reaches significant levels for downhill walking (Fig. 2). Thus, the tibiofemoral joint compressive forces and shear forces during level walking will approximate the calculated values, while for downhill walking the flexor moment produced by the hamstrings and gastrocnemius muscles will produce higher maximal stress levels in the tibiofemoral joints than those predicted here.

Morrison [13] calculated the knee joint shear forces and femorotibial joint compressive forces for different activities such as level walking, stair climbing and downhill walking. He too found much higher shear forces during downhill walking compared to the other activities. The present study indicated a shear force of 0.7–0.9 BW for level walking and 1.2–1.7 BW for downhill walking. These peak forces occurred in both studies during the early stance phase. The results of our study do, however, indicate higher values for both tasks. This can partially be explained by differences in cadence, as mentioned above, and the potential bias with a small sample of subjects, given the extremely high variability in kinetic data often reported [9, 23]. Furthermore, Morrison used a model that did not account for the rolling and gliding of the tibia on the femur, whereas the more sophisticated knee joint model from Nisell used in the present study not only accounted for these effects but has also been validated [15]. Moreover, comparisons between present results and those based on models using magnetic resonance imaging [21] show good agreement for level walking. Thus, downhill walking must be confidently stated to be a stressful task given the anterior shear forces in the order of 1.2–1.7 BW.

Complete failure of an ACL specimen obtained from a younger human reportedly occurs at stress levels between 1725 N [16] and 2160 N [24], with the failure point decreasing with increasing age [24]. Calculations of the anterior shear force for a 70-kg subject for downhill walking produces an anterior shear force of about 1000 N. This value is already close to the ultimate tensile strength of the ACL for a younger person and would lead to a complete failure in an older person. That this does not readily occur is obviously because joint stability is not only dependent on passive restraints but also on joint geometry, friction, and load as well as on muscular stabilisation [6, 7]. Of all these factors, the load imposed on the joint by muscular actions is one of the most important [7, 11]. In an in vitro study Markolf et al. [12] demonstrated that an application of 925 N of tibiofemoral contact force reduced the stress in the ACL by 36% at full extension and 46% at 20° flexion. In this study the load imposed on the tibiofemoral joint was shown to be 3.4–4 BW for level walking and 7.1–8.5 BW for downhill walking. It appears that this load sufficiently stabilises the knee joint during level walking where smaller shear forces occur and no further stabilisation from the hamstring muscles is necessary, as evidenced by the EMG results of the present study. On the other hand, the shear forces for downhill walking are potentially quite destabilising, and, despite an increased tibiofemoral load of 7–8.5 BW, further muscular stabilisation is necessary. Both Tokuhiro et al. [20] and the present study have shown prolonged EMG activity of the hamstring muscles into early stance and increased activity again during late stance for downhill walking. The hamstrings muscles, therefore, can be seen to be an important stabiliser of the knee joint and protector of the ACL during downhill walking. The concomitant activation of the hamstrings during quadriceps muscle activity (Fig. 2) will exert a posterior draw and thereby protect the ACL.

Draganich et al. [4] and Baratta et al. [2] have reported co-activation of the hamstrings and the quadriceps in healthy patients during the terminal phase of knee extension in slow isokinetic exercise, further supporting the hypothesis that co-activation of the hamstrings is an important stabilising factor with respect to the knee joint as well as a dynamic synergist of the ACL, even in normal subjects. The use of the semitendinosus tendon as a graft for an ACL reconstruction is therefore of dubious merit [5].

A further muscle that appears to stabilise the knee joint in early stance phase during downhill walking was found to be the gastrocnemius muscle. The EMG records showed two episodes of peak activity, one occurring at heelstrike and the other during late stance. While the peak during the late stance phase serves to plantarflex the foot for the push-off, there is no power generation necessary at the ankle joint at heelstrike. This first peak therefore suggests a stabilising function of the gastrocnemius muscle for the knee and ankle joints. The greatest gastrocnemius EMG activity occurred around heelstrike, which, when the electromechanical delay is taken into account, suggests that the maximal force production of the gastrocnemius muscle would be in the early stance phase and at a time when the maximal shear force occurred too. Mann and Inman [10] have suggested that when walking down a slope, the body requires the foot to be a rigid lever arm early in the walking cycle to counteract the moment created by the accelerating body. Moreover, early muscular action was considered to stabilise the knee joint during heel contact. Tokuhiro et al. [20] collected the EMG of the gastrocnemius and other muscles during different gradients of slope walking. They found that the gastrocnemius muslce acted earlier in downslope walking than in level walking, appearing to be already active at heel contact if the slope was more than -3°. These results are consistent with those reported here.

For reconstruction of the ACL-deficient knee most surgeons use a bone-patellar tendon – bone autogenous graft because of its strength, its ready availability and the ease with which immediate strong fixation can be obtained. This allows a postoperative treatment with early motion, full extension and early full bearing [18]. However, despite the initial strength of the patellar ligament graft, there is a graft pull-out strength as low as 200–500 N, depending on the fixation method used [8]. This will become the limiting factor during the early postoperative phase. The graft will subsequently undergo a process of ischaemic necrosis, revascularisation, proliferation and finally remodelling within 12 months, during which time the tensile strength of the graft progressively increases from a low of 26% at 3 months to 52% at 1 year, at least in the rhesus monkey [3]. In an arthroscopic and histological study of the remodelling process of allogenic tendon graft in the human knee conducted by Shino et al. [19] it was concluded that a graft reaches maturity not before 12–18 months in humans. It is therefore advisable to protect the graft from high stress levels during the postoperative period when the graft undergoes remodelling. However, complete or partial stress shielding can produce a hypertrophic and mechanically weak graft [25]. This suggests a postoperative care dilemma, where too much stress could endanger the graft and too little stress could end in a hypertrophic and weak graft. It is therefore extremely important to find the correct postoperative rehabilitation scheme for a good remodelling process of the graft. Insofar as level walking produces smaller shear forces and no co-activation of agonist and antagonist is necessary for the stabilisation of the knee joint, an accelerated postoperative rehabilitation including full weight bearing can be recommended without putting too much stress on the graft. However, a very different picture occurs for downhill walking. This task produces higher levels of anterior shear force and even healthy subjects are dependent on a complicated muscular activation pattern for the protection of the ligaments, necessitating co-activation of the hamstrings, gastrocnemius and quadriceps muscles during early stance. Since there may well be some muscular impairment postoperatively due to disturbed proprioception, an ACL graft could be endangered. Downhill walking should therefore be avoided during the postoperative phase in order to protect the reconstruction.

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