FORCES TRANSMITTED IN THE KNEE JOINT DURING STAIR ASCENT AND DESCENT

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ABSTRACT

The normal function of the knee relies upon a complicated and delicate mechanical interaction between the force-bearing structures, including muscles, ligaments and articular surfaces. Injury or damage to any of these structures will lead to the degradation or loss of joint function. Stair locomotion is one of the most common forms of motion in daily living. The purposes of the study were to analyze forces transmitted by the force-bearing structures of the knee during normal stair ascent and descent and to evaluate their mechanical differences. Compared to stair descent, higher peak patellar tendon forces, peak flexor forces, posterior cruciate ligament (PCL) forces and contact forces were required during stair ascent. In contrast, the anterior cruciate ligament (ACL) trended to bear higher forces during stair descent. It is suggested that one should be cautious in the use of stair ascent and descent as a rehabilitation exercise for patients with injuries or diseases of the cruciate ligaments and articular surfaces. The results of the study will be helpful for the understanding of the normal mechanics of the knee during stair locomotion, contributing to the design and evaluation of rehabilitation programs for patients with knee problems such as ACL injury and osteoarthritis.

Keywords : Knee Mechanics, Muscle Force, Ligament Force, Stair Activities, Locomotion.

1. INTRODUCTION

The normal function of the knee joint relies upon a complicated and delicate mechanical interaction between the force-bearing structures, including muscles, ligaments and articular surfaces. The ligaments together with articular surfaces control the passive motion of the joint while the dynamic stability of the joint requires additional muscular actions [1]. Injury or damage to any of these force-bearing structures will lead to degradation or loss of joint function. For instance, injury or rupture of the anterior cruciate ligament (ACL) will result in joint instability in the anterioposterial direction. Damage of the articular surfaces such as that in osteoarthritis (OA) will increase the friction between the articular surfaces and thus restrict the motion of the joint with combined pain. Therefore, knowledge of the biomechanical interactions of the force-bearing structures will be helpful for the design and evaluation of treatment of relevant diseases.

Stair locomotion is one of the most common forms of motion in daily living. A recent cardiopulmonary study showed that stair ascent required much more energy and increased heart rate compared to stair descent or level walking [2]. Regular stair-climbing exercises have been shown to improve cardiovascular fitness, reduce cholesterol levels, decrease body fat and increase the strength of the lower limb muscles [3~5]. On the other hand, it is generally assumed that stair descent loads the joints of the lower limb much more than stair ascent. Thus, patients have been suggested to avoid stair descent during the course of rehabilitation and thus step-ups are used in clinics for training. Since both stair ascent and descent are frequent activities of daily living, it seems necessary to have scientific evidence as to whether stair descent loads the lower limb joints more than stair ascent.

Forces transmitted by the force-bearing structures of the musculoskeletal system in living subjects have been estimated indirectly using cadavers or non-invasive modeling techniques because technological restrictions and ethical considerations have prevented the direct measurement of these forces. Joint contact forces and ligament forces can be evaluated on cadaveric knees but effects of muscle actions during dynamic movements are difficult to simulate in an *in vitro* condition. Use of instrumented prostheses allows the direct measurement of forces in bones and joints [6] but can only apply to a limited number of subjects and thus the results may not be representative to other subjects. Mathematical

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modeling, in conjunction with non-invasive experimental measurements, can predict the forces transmitted by the various elements of the locomotor system [7~15]. With this approach, inverse dynamics analysis is used to obtain the joint resultant forces and moments that are further distributed to individual force-bearing structures, lines of action and lever arms of which are determined by a geometric model of the system. This is the so-called force-distribution problem [16]. Unfortunately, the solution of the force-distribution problem is difficult because of the high degree of redundancy of the system. In other words, the available equations of dynamic equilibrium are too few to allow the determination of all the forces simultaneously. There have been several different approaches proposed in the literature to resolve the problem, including the reduction method [11,12], optimization methods [9,10,14,15] and the DDOSC method [7,8,17]. The only work that was validated against data measured simultaneously using instrumented prosthesis in the same subject was that of Lu *et al.* [18].

Compared to level walking, biomechanical studies on stair ascent and descent are relatively few and only limited to reporting joint angles, forces and moments [19~21]. Taylor *et al*. [22] measured using telemetry the axial forces, torques and bending moments transmitted in the shaft of an instrumented distal femoral prosthesis during stair locomotion. Taylor *et al*. [23] predicted the tibio-femoral joint contact forces during walking and stair climbing. Costigan *et al*. [24] used a subject-specific knee model together with joint resultant forces and moments to estimate the joint contact forces. However, it is noted that no study has been published on the forces transmitted in the muscles, ligaments and joint articular surfaces during stair ascent and descent.

The purposes of the present study were thus to analyze forces transmitted by the force-bearing structures of the knee in normal subjects during stair ascent and descent and to evaluate the differences in the calculated forces between these two stair activities. It is hoped that the results of the study will be helpful for the understanding of the normal mechanics of the knee during stair locomotion, which in turn contributes to the design and evaluation of rehabilitation programs for patients with knee problems such as ACL injury and OA.

2. MATERIALS AND METHODS

2.1 Experimental Method

Ten normal adults (age: 20 ± 1.1 years, height: 164.8 ± 6.1cm, weight: 63.3 ± 14.3kg) without any history of lower limb disease participated in this study with informed consents. For each subject, a total of 31 infrared retroreflective markers were attached to specific bony landmarks on each limb for the description of the motion of the segments, including anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanter, mid-thigh, medial and lateral femoral epicondyles, tibial tuberosity, head of fibula, medial and lateral malleoli, calcaneus, navicular tuberosity and the base of the fifth metatarsal. The medial condylar markers were used only during static subject calibration. They were then removed during dynamic activities to avoid interfering with the motion. The subject was asked to climb a custom-made three-step stair (height: 18cm; depth: 46cm) first ascending and then descending with both legs tested (Fig. 1). A seven-camera motion analysis system (VICON 512, Oxford Metrics, U.K.) was used to measure the three-dimensional trajectories of the markers with a sampling rate of 60Hz. The ground reaction forces (GRF) were also measured simultaneously at a sampling rate of 240Hz with two force platforms (AMTI, Mass., U.S.A.), which served as the second step of the three-step stair (Fig. 1). The starting position of the subject was adjusted by the examiner so that the tested foot could place naturally on the force platform during tested activities. Activities of eight major muscles of the right limb were monitored with a 10-channel electromyography (EMG) system (MA300, Motion Lab., U.S.A.), namely gluteus maximus, rectus femoris, vastus medialis, vastus lateralis, medial hamstrings, biceps femoris, tibialis anterior and medial gastrocnemius. The EMG of the muscles during maximum voluntary isometric contractions (MVIC) were also collected for each subject before stair locomotion tests. Raw EMG data were first rectified and then low pass filtered with a zero-lag fourth-order Butterworth filter with a cutoff frequency of 5Hz to obtain the linear envelopes of the EMG data that were then used to indicate muscle on/off status. Motion data from at least 3 trials for each test were collected.

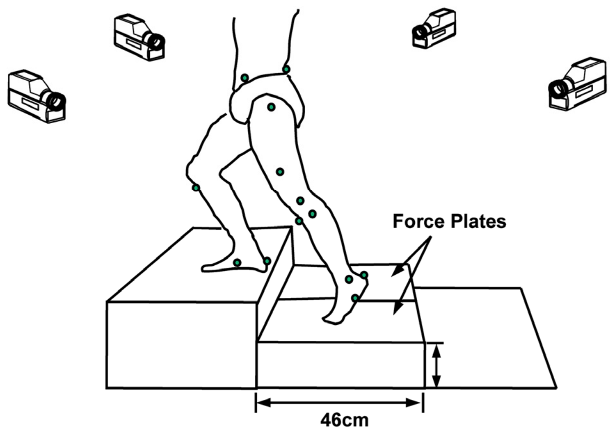

Fig. 1 Schematic diagram showing a subject ascending a three-step stair with the tested foot stepped on the second step fitted with two force platforms. Part of the markers placed on the bony landmarks for the tracking of the limb segments is also shown. A camera motion analysis system was used to measure the three-dimensional trajectories of the markers.

2.2 The Human Lower Limb Model

A validated model of the human locomotor system was adopted in the present study [18] for the analysis of the measured motion data. The human pelvis-leg apparatus was modelled in the sagittal plane as four rigid body segments, namely the pelvis, thigh, shank and foot, connected by model joints. The mobility of the tibiofemoral joint was controlled by a four-bar linkage, formed by the isometric fibers of the ACL and PCL as well as the lines joining their attachments on the femur and tibia (Fig. 2) [25,26]. Given the knee flexion angle, the model determines the orientations of the ligaments and the contact force. Formulae and model parameters describing the geometry of the linkage system were given in Zavatsky and O'Connor [27,28]. The knee extensor mechanism, composed of the patellofemoral joint, patellar and quadriceps tendons, was also included in the model. The patella was represented as a point, the intersection of the quadriceps and patellar tendons [26,29]. In conjunction with the 4-bar linkage model, the knee extensor mechanism model takes account of the rolling motion of the femur on the tibia, determining the change of geometry of the patellar tendon, quadriceps tendon and patellofemoral articular contact normal, throughout the full range of knee flexion. The ankle and hip were modelled as simple functional revolute joints, which is justified by previous *in vitro* and *in vivo* studies [30~33].

Eight muscles or muscle groups, iliopsoas, rectus femoris, vasti, hamstrings, gluteus maximus, gastrocnemius, soleus and tibialis anterior were incorporated into the lower limb model, represented by single lines joining their origins and insertions, wrapping around the underlying bones when necessary. The model was customised to individual subjects using homogeneous scaling techniques suggested by Brand *et al.* [16] and White *et al.* [34]. Muscles that controlled the motion of the knee joint together with the ligaments are shown in Fig. 2.

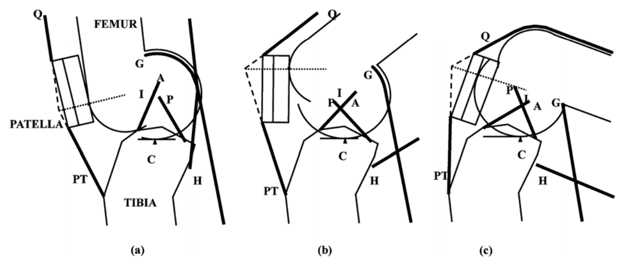

Fig. 2 The model tibiofemoral and patellofemoral joints with muscles and ligaments at (a) 0°, (b) 60°, (c) 120° flexion. Single thick lines are used to represent the force-bearing structures: ACL (A), PCL (P), patellar tendon (PT), quadriceps (Q), hamstrings (H), gastrocnemius (G). The tibiofemoral contact point (C) and instant center (I) moves posteriorly as the knee flexes.

2.3 Dynamic and Structural Analysis

The measured skin marker trajectories and force plate data were input into the model to calculate the intersegmental resultant forces and moments at the joint center during stance phase of stair locomotion using inverse dynamics. Inertial properties of the segments were determined using Dempster's coefficients [35]. Skin marker movement artefacts were minimised using the Global Optimisation Method [17]. The calculated intersegmental resultant forces and moments were transformed to the body-embedded coordinate systems of the distal segments and used for subsequent calculation of the forces transmitted in the muscles, ligaments and articular surfaces.

Since lines of action of muscles, ligaments and the tibiofemoral contact force were determined from the geometric model, only force magnitudes were considered as system unknowns except for the hip and ankle where the joint contact forces were each defined by two parameters: force magnitude and direction. Six unknowns for six structural members were considered at the knee: Q = quadriceps, H = hamstrings, G = gastrocnemius, A = ACL, P = PCL, and C = magnitude of the tibio-femoral contact force. Considering force and moment equipollence at knee joint, a total of 3 equations were formulated in terms of the 6 model unknowns, resulting in an indeterminate problem with infinite number of solutions.

To resolve the indeterminate problem, the Dynamically Determinate One-Sided Constrained (DDOSC) method was used [18]. The DDOSC method decomposes the original indeterminate problem into a series of dynamically determinate (DD) sub-problems established by considering at any one time only the number of unknowns (3) that makes the original problem determinate, setting the values of other forces to zero. Solutions corresponding to the DD sub-problems are referred to as DD solutions. Since muscles and ligaments transmit only tensile forces and joint articular surfaces resist only compression, DD solutions that violate these one-sided constraints are rejected. The remaining DD solutions are referred to as DDOSC solutions [18]. A total of 20 DD sub-problems for the knee joint were considered in the present study. The capability of the model to predict physiologically feasible solutions with the DDOSC method is examined by comparing the solutions with EMG evidence. In the present study, for any one instance where more than one DDOSC solution existed, a weighted combination of the DDOSC solutions that minimized the sum of squared muscle forces was selected. All the calculated forces were normalized to body weight (BW) and moments to body weight and leg length (BWLL).

2.4 Statistical Analysis

The curves of the calculated variables during the stance phase, including resultant forces and moments and force values of the muscles, ligaments and contact forces, for all the subjects were ensemble-averaged to provide general patterns of the forces and moments.

Peak values of these variables were compared between stair ascending and stair descending using paired-t test (SPSS, U.S.A.), with a significance level of 0.05.

3. RESULTS AND DISCUSSION

The stance phase (SP) of ascent can be divided into three subphases: initial double stance, single stance and terminal double stance [36]. Initial double stance phase (0 ~ 25% SP) begins at heelstrike and ends at contralateral toe-off, during which the main task of the limb is weight acceptance. Single stance phase (25 ~ 75% SP) begins at contralateral toe-off and ends at contralateral heelstrike on the upper step. The main task of this period is to push the body forwards and upwards to the next step. Terminal double stance phase (75 ~ 100% SP) follows single stance phase and ends at ipsilateral toe-off. During this period, the body weight is released to the contralateral limb. Similar phases are also present during stair descent except that during single stance phase the main task of the supporting limb is to lower the body. In the present study, only results for the knee joint are presented in the following with reference to the subphases described.

3.1 Intersegmental Forces and Moments

During stair ascent, the knee was at about 60° flexion at initial contact and remained so during the first 15% SP (Fig. 3(a)). During 0 ~ 25% SP, the main task of the knee joint was to accept the body weight released from the contralateral side, with a rapid increase in the distally directed force component applied to the shank (Fig. 3(c)). After weight acceptance, increasing anteriorly directed force components with a maximum of 0.42BW and increasing extensor moments with a maximum of 0.161 BWLL were necessary to push the body forward and upward (Figs. 3(b) and 3(d)). During this period, the knee extended gradually from 60° flexion to a minimum of about 15° and remained almost unchanged for the rest of the stance phase (Fig. 3(a)). With the knee and leg at an almost extended position during 50 ~ 100% SP, the ground reaction forces passed close to the knee joint center, reducing the moments required for the muscles (Fig. 3(d)). Overall, the knee moment curve had only one peak (0.161BWLL at about 25% SP) over the entire stance phase. Big extensor moments at the knee were required during 0 ~ 60% SP while relatively smaller flexor moments for the rest of the phase. The anterior component of the intersegmental force at the knee had two peaks: 0.42BW at about 25% SP and 0.37BW at about 85% SP (Fig. 3(b)). There was no significant difference between the two peaks ($p > 0.05$). The distal component also had two peaks: 0.93BW at about 25% SP and 1.16BW at about 85% SP (Fig. 3(c)), the later being significantly bigger than the former ($p < 0.05$).

During stair descent, the knee flexed slowly before 25% SP with small flexor moment generated during 0 ~ 15% SP. The knee was at about 9° flexion at heelstrike and flexed slowly to reach 23° at about 25% SP (Fig. 3(a)). During this period, the main task of the knee joint was to accept the body weight shifted from the contralateral side and to absorb impact loadings due to the lowering of the body, with a rapid increase in the distally and anteriorly directed force components as well as extensor moments (Fig. 3(c)). During 25 ~ 50% SP, the knee remained at about 20° of flexion. From 50% SP, the knee began to flex rapidly and reached a maximum of more than 60° at the end of the stance phase. Similar to stair ascent, the ground reaction forces were directed posteriorly during the whole period of stance phase, resulting in anteriorly directed force components at the knee (Fig. 3(b)). The knee moment curve had two peaks: 0.096BWLL and 0.13BWLL at 25% SP and 85% SP, respectively. There was no significant difference between the two peaks ($p > 0.05$). The anteriorly directed force components had also two peaks: 0.41BW at about 25% SP and 0.57BW at about 85% SP. No significant difference existed between the two peaks ($p > 0.05$). The distal component also had two peaks: 1.26BW at about 20% SP and 0.72BW at about 80% SP (Fig. 3(c)), the former being significantly bigger than the later ($p < 0.01$).

The intersegmental resultant forces and moments at the knee in the sagittal plane reported in this study agreed well with those in previous studies [19,20,24]. In comparison of the mean forces and moments between stair ascent and descent, the second peak of the anteriorly directed force components during stair descent was found to be significantly higher than that during stair ascent ($p < 0.05$), while no significant difference between the first peaks. The first peak of the distal force components during ascent was higher than that during descent ($p < 0.05$) while the opposite was found for the second peak. The peak extensor moment during stair

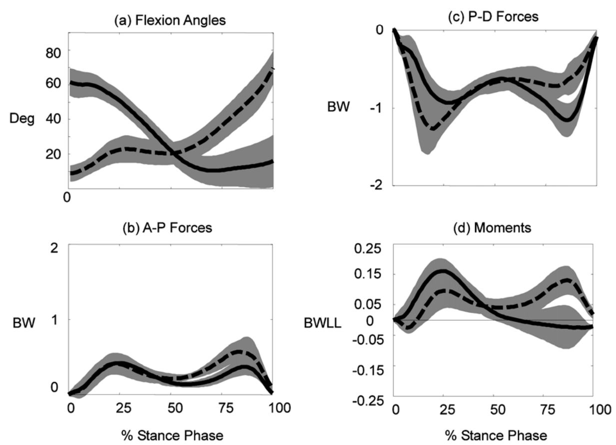

Fig. 3 (a) Joint flexion angles, (b) anterior-posterior (A-P) resultant forces, (c) proximal-distal (P-D) resultant forces and (d) resultant moments at the knee during stair ascent (solid lines) and stair descent (dashed lines). Shadow areas are one standard deviation from the mean curves. (BW: body weight; BWLL: body weight times leg length) ascent was higher than both peaks during descent ($p < 0.05$), indicating that greater moment was required during ascent.

3.2 EMG Activities and Forces in the Knee

The muscle activities indicated by EMG during stance phase are shown in Fig. 4. The ensemble-averaged muscular, ligamentous and contact forces during stance phase of stair ascent and descent are shown in Fig. 5. Good agreement in activation patterns between the calculated muscle forces and corresponding EMG was found. The times when the peak muscle forces occurred were also in agreement with those of the corresponding peak EMG (Figs. 4 and 5), considering the constant lag of about 30ms between the onset of EMG and muscle force generation [37]. There was also a lag of more than 30 ms for muscle forces to diminish after the end of muscle EMG activity. This lag for quadriceps was reported to be around 250 ~ 300ms [38]. Therefore, a muscle can still generate continuous force even though there is a short silence between two contractions.

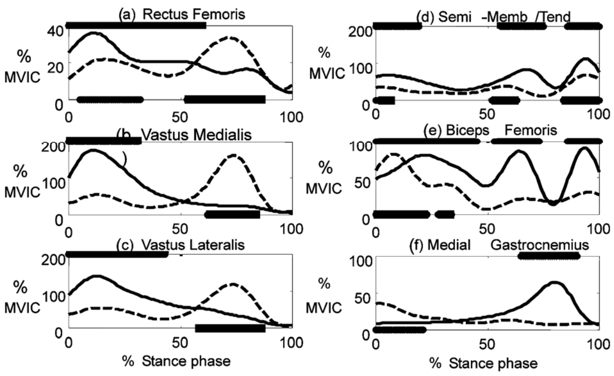

Fig. 4 Ensemble-averaged EMG linear envelops during stance phase of stair ascent (solid lines) and descent (dashed lines). The phasic activity of the muscles are shown as thick solid bars at the top for stair ascent and bottom for stair descent

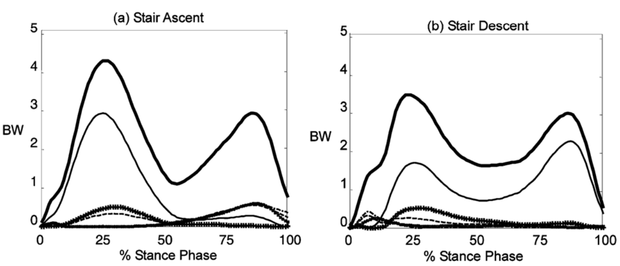

Fig. 5 The mean forces transmitted by the patellar tendon (thin solid line), hamstrings (dashed line), gastrocnemius (dot and dashed line), ACL (+), PCL (x) and tibiofemoral articular surfaces (thick solid line) during the stance phase of (a) stair ascent and (b) descent. (BW: body weight)

3.3 Forces in the Knee during Stair Ascent

During weight acceptance (0 ~ 25% SP) of stair ascent, the quadriceps was active to provide extensor moments for the knee joint. Quadriceps forces built up quickly during this period and reached a peak of 2.94BW at around 25% SP, Fig. 5(a). Antagonistic hamstrings activity was also found during this period, with the first peak of 0.08BW at about 5% SP (Fig. 5(a)). A small burst of gastrocnemius force of about 0.12 BW was also present (Fig. 5(a)). At this moment, the PCL was found to transmit a peak force of 0.08BW in resisting the posterior drawing force. The hamstrings, gastrocnemius and PCL forces together contributed to the initial increase of the joint contact force (Fig. 5(a)). The ACL force was relatively small during this period. After 5% SP, the quadriceps force increased gradually and reached a peak of 2.94BW to generate the peak extensor moment (0.161BWLL) at about 25% SP. During 5 ~ 25% SP, the hamstrings and ACL worked together to balance the anterior drawing force produced by the quadriceps, reaching their peak values of 0.34 and 0.51BW at about 30% SP, respectively. The vertical components of these forces together contributed to the first peak of the contact force (4.31BW) around 25% SP. The gastrocnemius and PCL were both resting between 5 ~ 25% SP. From 25% to 55% SP, the knee extended continuously with the ground reaction force drawn closer to the knee joint center, resulting in decreased intersegmental forces and moments. These reduced loading caused the decrease in the quadriceps, hamstrings and ACL forces, which further led to the decrease of the magnitude of the contact force to a minimum of 1.11BW. For the rest of the stance phase after 55% SP, quadriceps maintained a constant force of about 0.25BW (0.18 ~ 0.28BW) while both gastrocnemius and hamstrings increased their activity to produce forces up to 0.56 and 0.58BW respectively and generate negative moments at the knee. Increased gastrocnemius and hamstrings forces were needed mainly to extend the hip and plantarflexed the ankle for the propulsion of the body forward and upward to the next step. Since both the muscles were biarticular, these requirements inevitably imposed antagonistic moments at the knee. Antagonistic moments have been suggested to increase the joint stiffness so this arrangement might help ensure the dynamic stability of the knee during the propulsion phase of stair ascent. With the increase of hamstrings and gastrocnemius forces, the PCL and contact forces both increased to 0.60BW and 2.95BW, respectively, while little force was found to be transmitted in the ACL (less than 0.06BW) after 55% SP.

3.4 Forces in the Knee during Stair Descent

During early weight acceptance of stair descent (0 ~ 15% SP), the hamstrings and gastrocnemius generated forces of up to 0.32 and 0.43BW, respectively (Fig. 5(b)) to provide the required flexor moment (Fig. 3(d)). At this moment, since both muscles were pulling posteriorly, the PCL was found to transmit a peak force of 0.26BW to stabilize the knee. The hamstrings, gastrocnemius and PCL together provided a net anterior shear force to counter-balance the external posterior shear force. These force-bearing structures together also contributed to the initial increase of the joint contact force (Fig. 5(b)). During this period, little quadriceps and ACL forces were found (Fig. 5(b)). During 15 ~ 25% SP, the quadriceps force increased gradually and reached a peak of 1.71BW around 25% SP, generating the first peak extensor moment (0.096BWLL). During this period, the hamstrings and ACL worked together to balance the anterior shear force produced by quadriceps, reaching their peak values of 0.27 and 0.52BW around 25% SP, respectively. The vertical components of these forces together contributed to the first peak of the contact force, with a peak magnitude of 3.49BW at 25% SP. During this period, the gastrocnemius and PCL forces were both less than 0.3BW.

During 25 ~ 55% SP, the knee remained flexed at about 23° and co-contraction of the quadriceps and hamstrings was found. With the extensor moments decreasing, the quadriceps and hamstrings forces decreased continuously from their peaks of 1.71 and 0.27BW to about 0.7 and 0.1BW, respectively. This co-contraction was to ensure the stability of the knee joint during the lowering of the body center of mass due to the downward swing of the contralateral limb from the upper step. During this period, the ACL force decreased continuously from 0.52 to 0.2BW while the PCL force remained almost constant around 0.05 BW. The contact force also decreased from its peak value of 3.49BW to a minimum of 1.7BW. From 55% SP, the quadriceps force began to increase and reached a maximum of 2.27BW around 88% SP. As the quadriceps force increased, the hamstrings and gastrocnemius forces decreased, giving a higher extensor moment to decelerate the lowering of the body. During this period, the ACL force remained almost constant around 0.2BW in order to assist the anterior shear forces applied to the tibia. Although the peak quadriceps force was much higher than that during early single stance, its anterior shear component was less because the knee was at higher flexion (about 55°) and the patellar tendon almost vertical to the tibial plateau. The contact force reached a peak of 3.01BW at about 87% SP. Little forces were transmitted by the gastrocnemius and PCL.

It is noted that an increase in quadriceps force often led to increase in the ACL force, especially at a lower knee flexion angles (Figs. 5 and 6). In contrast, an increase in hamstrings or gastrocnemius forces often led to increase in the PCL force. Co-contractions of the quadriceps and hamstrings were found in the early half of stance phase during both stair ascent and descent, when the dynamic stability of the knee was essential. However, these increases of muscle forces, especially under co-contraction, always gave rise to high contact forces.

3.5 Comparisons of the Forces in the Knee between Stair Ascent and Descent

Comparisons of the forces transmitted in the individual force-bearing structures between stair ascent and descent are given in Fig. 6. The quadriceps force had a high peak at early single stance phase to extend the knee from a high flexed position to elevate the body upward and forward to the next step. This peak was found to be much bigger than the two peaks during stair descent ($p = 0.002$ for the first peak and $p = 0.008$ for the second), suggesting that the effort required for the quadriceps to ascend was higher than to descend. Generally, the isolated contraction of the quadriceps would load the ACL at knee flexion angles less than 70 ~ 100° but load the PCL at flexion angles higher than 100° since the patellar tendon is almost vertical to the tibial plateau around 70 ~ 100° flexion [39]. In the present study, the ACL was found to resist the anterior shear component of the quadriceps force only during the first half of stance both during ascent and descent. The quadriceps did not load the ACL much during the later half of stance during stair descent because of the higher knee flexion angle. During ascent the quadriceps exerted large forces at about 60° of knee flexion while smaller quadriceps forces were required for the knee at about 20° of flexion. Since the shear component of the quadriceps force during descent was much bigger than that for ascent, the peak values of the ACL forces during descent were bigger than those during ascent in most of the cases although has not yet reached statistical significance ($p = 0.086$). It seems that stair descent may have greater risk in damaging the ACL than stair ascent,

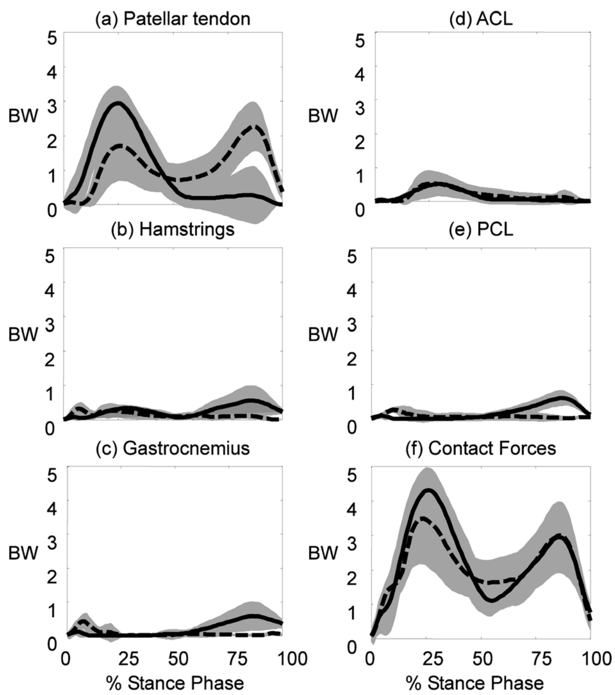

Fig. 6 The mean forces transmitted by the (a) patellar tendon, (b) hamstrings, (c) gastrocnemius, (d) ACL, (e) PCL and (f) tibiofemoral articular surfaces during stair ascent (solid lines) and descent (thin dashed lines). Shadow areas are one standard deviation from the mean curves. (BW: body weight)

which is of significant clinical relevance for patients with ACL deficiency and/or reconstruction. However, further study with bigger subject sample size may be needed to confirm further these findings.

The EMG showed that the hamstrings were active to provide flexor moment at the knee during most of the stance phase of stair ascent, especially the biceps, Fig. 4. The corresponding hamstrings forces demonstrated a three-peak pattern: one immediately after heelstrike, one during early single stance and the other at the middle of terminal double stance. The third peak was higher than the other two, suggesting that higher hamstring effort was needed for propulsion during terminal double stance of ascent, Fig. 6(b). On the contrary, higher hamstring forces were needed during initial double stance and early single stance for stair descent, Figs. 4 and 6(b). The hamstrings generated a peak force immediately after heelstrike and then remained active and exerting decreasing force throughout the rest of stance phase. Compared to stair ascent, the mean peak hamstring force immediately after heelstrike was higher during stair descent as higher flexor moment was needed (Fig. 3(d)), although has not yet reached statistical significant level ($p = 0.057$). The gastrocnemius and PCL forces both showed similar two-peak patterns during ascent and one-peak patterns during descent, Fig. 6. Similar to hamstrings forces, both the mean peak forces of gastrocnemius and PCL immediately after heelstrike were higher during descent with significance found only in the PCL forces ($p = 0.110$ for gastrocnemius and $p < 0.01$ for PCL, respectively). However, peak gastrocnemius and PCL forces around the middle of terminal double stance phase during ascent were higher than the peak forces during descent ($p < 0.05$), Fig. 6(e).

There were two peaks in the calculated tibiofemoral contact forces during both stair ascent and descent, Fig 6(f). No significant difference was found for the second peak contact forces between stair ascent and descent ($p = 0.711$). The first peak of the mean contact force during stair ascent was higher than that during descent but no statistical significance was obtained. The first peak was also found to be higher than the second peak during stair ascent ($p = 0.002$) but no significant difference was found for stair descent ($p = 0.219$). This indicates that the knee joint needed to bear higher force in the early stance phase of stair ascent when the knee started to extend. This may be a potential risk of joint contact surface damage that could facilitate the development of osteoarthritis.

The differences of patterns and peak values in the forces transmitted in the force-bearing structures between ascent and descent indicated that different strategies of recruitment of these force-bearing structures were adopted for stair ascent and descent. The major differences existed during the initial and terminal double stance phases. During stair ascent, the leading limb accepted the body weight from the trailing limb before it started to extend to lift the body upward to be above the upper step. During descending, while the leading limb was accepting the body weight from the trailing limb, the knee and ankle joints were flexing to have the body move forward and downward to be above the lower step.

3.6 Comparisons of the Knee Forces between Stair Activities and Level Walking

Since data available in the literature on the forces transmitted by the muscles, ligaments and articular forces at the knee are mainly for level walking, the calculated forces transmitted by these force-bearing structures during stair locomotion are compared to those data. Several studies estimated ACL forces during level walking [7,8,11,40]. Compared to the reported peak ACL forces, ranging from about 0.2BW [11] to 3.5BW [7], the peak ACL force calculated in the present study during stair ascent (0.51BW) and descent (0.52BW) lied in between. This suggests that the peak loading for the ACL during stair locomotion may not be significantly different from those during level walking as one would normally assumed.

The reported PCL peak forces during level walking were as diverse as those for the ACL. Morrison [11] reported that the PCL transmitted 0.4BW during the final 45% of stance phase while the PCL was found to be almost unloaded in Shelburne *et al.* [40]). In the present study, the mean peak PCL forces were 0.6 BW during stair ascent and 0.26BW during descent, suggesting that the PCL may experience higher loading during stair locomotion than in level walking. However, since the calculated ACL and PCL forces are much less than their normal ultimate strength (both over 1500N [41~43]), this implies that stair activities are safe to healthy knee ligaments.

The peak quadriceps forces during stair ascent (2.94BW) and descent (1.71 and 2.27BW) were much higher than the maximal forces during level walking (about 1BW [11] and 1.7BW [40]) as expected. During stair ascent, the predicted peak hamstrings forces (0.58 BW) were bigger than level walking according to Shelburne *et al.* [40] (less than 0.35BW) but smaller than that reported by Morrison [11] (about 1.6BW during early stance phase). The peak hamstrings forces during stair descent (0.32BW) were similar to those of level walking (0.35BW reported by Shelburne *et al.* [40]) but much lower than the 1.6BW reported by Morrison [11]. On the other hand, peak gastrocnemius forces during stair ascent (0.56BW) and descent (0.1BW) were both lower than those during level walking (about 1.3BW [11] and 1.2BW [40]). It is noted that Morrison [11] did not consider hamstrings and gastrocnemius co-contractions, the reported peak force values for these two muscles might be overestimation.

Morrison [11] and Taylor *et al.* [23] calculated mean peak tibio-femoral contact forces as 3.03 and 3.1BW respectively during level walking. Taylor *et al.* [23] also predicted a peak contact force of 5.4BW during the early stance phase of stair ascent. The mean peak contact forces obtained from the present study was 4.3BW during stair ascent and 3.49BW during stair descent, both higher than those during level walking.

The difference between the peak contact forces during stair ascent reported by Taylor *et al.* [23] and the present study may be due to different stair design. The stair in Taylor *et al.* appeared to be higher.

4. CONCLUSIONS

Forces transmitted by the force-bearing structures in the knee joint during stair locomotion were analyzed using a validated lower limb model. The knee extensors, the quadriceps, were found to transmit significantly higher peak forces during stair ascent than during stair descent. The knee flexors, hamstrings and gastrocnemius, were activated mainly during the late stance phase of stair ascent, generating big forces for pushing the body forward and upward through the extension of the hip and ankle. During stair descent, the knee flexors were activated mainly during early stance phase for weight acceptance with peak forces smaller than those during stair ascent. Both the ACL and PCL were loaded during the stair activities. The ACL resisted the anterior shear force only during the first half of stance both during ascent and descent. No significant differences were found in the forces transmitted by the ACL between the stair activities but the ACL tended to transmit higher force during stair descent. The PCL were loaded in response to the net posterior shear force components of the muscle actions and external forces. Compared to stair descent, higher PCL forces were required during stair ascent. The peak tibio-femoral contact forces during stair ascent and descent were not significantly different. Nonetheless, there was a noted trend for the knee joint to bear higher mean peak contact forces during stair ascent. Peak contact forces during both stair activities were bigger than those during level walking. From the results of the present study, it is suggested that one should be cautious in the use of stair ascent and descent as a rehabilitation exercise for patients with injuries or diseases of the cruciate ligaments and knee OA.

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