Clinical Biomechanics 28 (2013) 98–103

Contents lists available at SciVerse ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Hip and knee joint loading during vertical jumping and push jerking

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ARTICLE INFO

Article history:
Received 2 November 2011
Accepted 18 October 2012

Keywords:
Musculoskeletal modeling
Inverse dynamics
Joint contact forces
Vertical jumping
Weightlifting

ABSTRACT

Background: The internal joint contact forces experienced at the lower limb have been frequently studied in activities of daily living and rehabilitation activities. In contrast, the forces experienced during more dynamic activities are not well understood, and those studies that do exist suggest very high degrees of joint loading.
Methods: In this study a biomechanical model of the right lower limb was used to calculate the internal joint forces experienced by the lower limb during vertical jumping, landing and push jerking (an explosive exercise derived from the sport of Olympic weightlifting), with a particular emphasis on the forces experienced by the knee.
Findings: The knee experienced mean peak loadings of 2.4–4.6 × body weight at the patellofemoral joint, 6.9–9.0 × body weight at the tibiofemoral joint, 0.3–1.4 × body weight anterior tibial shear and 1.0–3.1 × body weight posterior tibial shear. The hip experienced a mean peak loading of 5.5–8.4 × body weight and the ankle 8.9–10.0 × body weight.
Interpretation: The magnitudes of the total (resultant) joint contact forces at the patellofemoral joint, tibiofemoral joint and hip are greater than those reported in activities of daily living and less dynamic rehabilitation exercises. The information in this study is of importance for medical professionals, coaches and biomedical researchers in improving the understanding of acute and chronic injuries, understanding the performance of prosthetic implants and materials, evaluating the appropriateness of jumping and weightlifting for patient populations and informing the training programmes of healthy populations.

1. Introduction

The quantification of the forces experienced by the hip and knee during movement has been of great interest to the biomedical research community and there have been a large number of studies that have sought to quantify this loading through both musculoskeletal modeling techniques and direct measurement. The majority of these studies have focussed on activities of daily living (ADLs; movements like "sit to stand", "stand to sit", gait, stair ascent/descent), or rehabilitation exercises characterized by relatively slow execution speeds (exercises like the squat or lunge). The breadth of this literature, allows a typical, albeit quite wide, range for the loading during these types of activities to be suggested. For instance, at least 15 different groups have calculated internal knee forces during squatting using musculoskeletal modeling techniques (Collins, 1994; Dahlkvist et al., 1982; Escamilla et al., 1998; Nagura et al., 2006; Nisell, 1985; Reilly and Martens, 1972; Salem and Powers, 2001; Sharma et al., 2008; Shelburne and Pandy, 1998, 2002; Smith et al., 2008; Thambyah, 2008; Toutoungi et al., 2000; Wallace et al., 2002; Wilk et al., 1996) and the internal forces suggested during body weight squatting include a patellofemoral joint force (PFJF) range of 2.5–7.6 × BW and a tibiofemoral joint force (TFJF) range of 2.5–7.3 × BW.

In contrast, there are fewer musculoskeletal modeling studies that have sought to understand the loading of the hip and knee joints during more dynamic movements with faster execution speeds. Those studies that do exist are often based on simple biomechanical models with inherently limiting assumptions and which thus may not accurately capture the nature of the joint loading (Nisell and Mizrahi, 1988; Simpson and Kanter, 1997; Simpson and Pettit, 1997; Smith, 1975). In particular, there is a tendency for these studies to report joint loadings that seem very high in comparison to those found in ADLs, even when accounting for a premium attributable to the more demanding nature of these activities. For example, Simpson and colleagues (Simpson and Kanter, 1997; Simpson and Pettit, 1997; Simpson et al., 1996) found that during a landing from a travelling jump (a horizontal jump to a single leg landing) the PFJF was 10.4 × BW and the TFJF 16.8 × BW. Similarly, in a pioneering study, Smith (1975) suggested that the TFJF experienced during a jump landing was in the range of 17.0–24.4 × BW. These high values may be a result of the lack of detail in the biomechanical models employed (Cleather and Bull, 2010b, 2012b) or even inaccurate model assumptions. In recent years, the prevalence of sporting injuries to the anterior cruciate ligament (ACL) of the knee (Majewski et al., 2006) has prompted an interest in quantifying the loading of this structure

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http://dx.doi.org/10.1016/j.clinbiomech.2012.10.006

during movement, also by employing musculoskeletal modeling techniques (Kernozek and Ragan, 2008; Pflum et al., 2004). A common approach is to calculate the anterior shear force (that is the force that tends to displace the tibia anteriorly on femur) and to use this as a proxy for the ACL loading (as the ACL is the primary restraint to anterior drawer of the knee). However, these studies also tend to be based upon inappropriately simple biomechanical models (Sell et al., 2007; Yu et al., 2006), and thus even a clear idea as to the shear forces experienced by the knee is largely unknown.

The development of instrumented prostheses has permitted the in vivo measurement of forces in the hip and knee, and provided new insights. For instance, D'Lima and colleagues have shown that during ADLs the magnitude of the TFJF is in the range of 2.0–3.0 × BW, but that during sporting activities (including jogging, tennis and golf) this rises to 3.0–4.5 × BW (D'Lima et al., 2005b, 2005a, 2006, 2007, 2008). These values also seem to suggest that the higher internal forces predicted during more dynamic activities by earlier biomechanical models could be questionable. The highly invasive nature of this research restricts these studies to patient populations (of often advanced ages) however, and it does seem likely that young, healthy populations might experience a greater loading.

It is clear that the magnitude of the forces experienced by the hip and knee joints during dynamic activities characterized by rapid movement speeds is not well understood. In particular, a typical upper range for the loading of the hip and knee joints in sporting movements in young healthy populations is generally unknown. The purpose of this study was therefore to use a previously developed model of the musculoskeletal model of the lower limb (Cleather and Bull, 2010b; Cleather et al., 2011a, 2011b) to quantify the nature and magnitude of the forces experienced at the joints of the lower extremity by a young athletic male population during vertical jumping and push jerking (two lower extremity activities characterized by high movement speeds and force loading and that are similar in kinematic character) with a particular focus on the forces experienced by the knee.

2. Methods

In this study a previously described biomechanical model (Cleather, 2010; Cleather and Bull, 2010a, 2010b; Cleather et al., 2011a, 2011b) of the right lower limb was employed to calculate the internal joint forces produced during vertical jumping and push jerking. The validation and verification of the model has been described in previous work (Cleather, 2010; Cleather and Bull, 2010b; Cleather et al., 2011a, 2011b) as has the sensitivity of the model to some key parameters (Cleather, 2010; Cleather and Bull, 2010a, 2010b, 2011). The study was approved by the local research ethics committee and all participants provided informed consent. Twelve athletic males (mean age 27.1 SD 4.3 years; mean mass 83.7 SD 9.9 kg) were recruited to take part in this study. After performing a standardized warm up consisting of lower extremity body weight exercises (such as squats, lunges and vertical jumps) each subject performed 5 maximal countermovement jumps with their hands on their hips and the highest jump (mean height 0.38 SD 0.05 m) was chosen for analysis. Nine of the subjects (mean age 27.3 SD 4.1 years; mean mass 84.1 SD 10.7 kg) who where familiar with the push jerk exercise (more than six months experience in Olympic weightlifting) also performed 3 repetitions of a push jerk with 40 kg — a movement derived from the competitive sport of Olympic weightlifting where a barbell is thrust overhead primarily by forces produced by extension of the lower limb joints. The data set comprised the position of reflective markers placed on key anatomical landmarks (Van Sint Jan, 2005; Van Sint Jan and Croce, 2005) determined using the Vicon motion capture system (Vicon MX System, Vicon Motion Systems Ltd, Oxford, UK) and the ground reaction force recorded by a portable force plate (Kistler Type 9286AA, Kistler Instrumente AG, Winterthur, Switzerland). The marker set employed in this study is described in detail elsewhere (Cleather, 2010), and comprises markers on the pelvis (4 markers on the anterior and posterior supra-iliac spines), thigh (5 markers — including markers on the medial and lateral epicondyles), calf (5 markers — including markers on the medial and lateral epicondyles) and foot (4 markers — including markers on the rear of calcaneus and the head of the second metatarsal). As the musculoskeletal model is of the right limb alone, each subject performed each trial with only their right foot on the force plate, thus the ground reaction force was that impressed by the right limb alone. All data were collected at 200 Hz. The raw data were filtered using generalized cross validatory spline filtering (Woltring, 1986; otherwise known as a Woltring filter) using a 5th order spline and a cut-off frequency of 10 Hz. Following the recommendation of Bisseling and Hof (2006), the force data were filtered using the same cut-off frequency as the kinematic data.

The musculoskeletal model consists of a linked series of four segments representing the foot, calf, thigh and pelvis articulated by ball and socket joints at the ankle, knee and hip. After filtering these segments were constructed from the positions of the markers using the method of Horn (1987) to establish the position and orientation of each segment. The anthropometry used in the model was taken from the work of de Leva (1996).

The data of Klein Horsman et al. (2007) were used to create a subject-specific musculoskeletal geometry of the lower limb. This consisted of 163 different line elements representing 38 different muscles of the lower limb. The position of the patella relative to the femur was calculated using the Klein Horsman data to determine the position of the patellar origin relative to the femur as a function of the knee flexion angle. The orientation of the patella relative to the femur (i.e. its sagittal plane rotation) for a given knee flexion angle was calculated using the data of Nha et al. (2008) using spline interpolation (using "Numerical Recipes in C++"; Press et al., 2002). The patellofemoral joint model also included the addition of via points to model the wrapping of the quadriceps around the femoral condyles in deep knee flexion. This was achieved by simply defining a via point for each quadriceps muscle element through which the element was constrained to pass once the quadriceps had begun to wrap around the femoral condyles. Finally, the changing ratio between quadriceps and patellar tendon forces (Mason et al., 2008) with increased knee flexion angle was calculated based upon the geometrical relationship between patella, patellar tendon and quadriceps tendons assuming the maintenance of force and moment equilibrium at the patella.

Muscle forces were determined using an optimization based approach to inverse dynamics (Cleather, 2010; Cleather et al., 2011a, 2011b). The inverse dynamics method of Dumas et al. (2004) was used to formulate the equations of motion of each segment as a function of the unknown muscle forces:

$$\begin{bmatrix} \hat{S}^t{}_i \\ \sum_{j=1}^{K} F_j \cdot \hat{r}_{ji} \times \hat{n}_{ji} - \sum_{j=1}^{K} F_j \cdot \hat{r}j(i-1) \times \hat{n}_{j(i-1)} + \sum_{j=1}^{K} b_{ji} F_j \cdot \hat{d}_i \times \hat{n}_{j(i-1)} \end{bmatrix} = \begin{bmatrix} m_i E_{3\times3} & 0_{3\times3} \\ {}_{mi}\tilde{c}_i & I_i \end{bmatrix} \begin{bmatrix} \hat{a}_i - \hat{g} \\ \ddot{\hat{\theta}}_i \end{bmatrix} + \begin{bmatrix} 0_{3\times1} \\ \dot{\hat{\theta}}_i \times I_i \dot{\hat{\theta}}_i \end{bmatrix} + \begin{bmatrix} E_{3\times3} & 0_{3\times3} \\ \tilde{d}_i & E_{3\times3} \end{bmatrix} \begin{bmatrix} \hat{S}_{i-1} \\ \hat{M}_{i-1} \end{bmatrix} \quad (1)$$

b_{ji} 1 for biarticular muscles that cross but do not attach to segment i;
b_{ji} 0 for all other muscles

Where $\hat{M}_{i-1}$ was set to zero for $i > 1$ and:

i segment number or joint number (1 represents the most distal segment or joint)
$\hat{S}_i$ proximal joint reaction forces
$\hat{S}_{i-1}$ distal joint reaction forces
$\hat{M}_{i-1}$ distal joint moments
I_i inertia tensor
$\dot{\hat{\theta}}_i$ angular velocity about COM