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The influence of simulation model complexity on the estimation of internal loading in gymnastics landings

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Abstract

Evaluating landing technique using a computer simulation model of a gymnast and landing mat could be a useful tool when attempting to assess injury risk. The aims of this study were: (1) to investigate whether a subject-specific torque-driven or a subject-specific muscle-driven model of a gymnast is better at matching experimental ground reaction forces and kinematics during gymnastics landings, (2) to calculate their respective simulation run times and (3) to determine what level of model complexity is required to assess injury risk. A subject-specific planar seven-link wobbling mass model of a gymnast and a multi-layer model of a landing mat were developed for this study. Subject-specific strength parameters were determined which defined the maximum voluntary torque/angle/angular velocity relationship about each joint. This relationship was also used to produce subject-specific 'lumped' muscle models for each joint. Kinetic and kinematic data were obtained during landings from backward and forward rotating gymnastics vaults. Both torque-driven and muscle-driven models were capable of producing simulated landings that matched the actual performances (with overall percentage differences between 10.1% and 18.2%). The torque-driven model underestimated the internal loading on joints and bones, resulting in joint reaction forces that were less than 50% of those calculated using the muscle-driven model. Simulation time increased from approximately 3 min (torque driven) to more than 10 min (muscle driven) as model complexity increased. The selection of a simulation model for assessing injury risk must consider the need for determining realistic internal forces as the priority despite increases in simulation run time.

Keywords: Optimisation; Forces; Impact; Muscle driven; Torque driven

1. Introduction

When landing from a dismount in Artistic Gymnastics, the aim is to reduce the velocity of the mass centre to zero while over the base of support using a single placement of the feet. Any steps or unsteadiness during landing can result in a score deduction between 0.1 and 0.3 (F.I.G., 2001). The successful landing from a vault (or other apparatus) poses a problem for a gymnast who must trade off technical difficulty with the probability of a successful landing and with the risk of injury. While minimising injury is a concern, it is not always of paramount importance in competitive environments, such as at the Atlanta 1996 Olympics Games, where an already injured gymnast completed a vault, further exacerbating her injury. The gymnast and coach aim to maximise performance and develop training regimes and technical strategies to achieve this. Similarly, most of the computer simulation research in gymnastics, jumping and landing has been aimed at improving performance (Hiley and Yeadon, 2003; King and Yeadon, 2004) and has not considered the potential injury risks. Using a gymnast to investigate different landing techniques experimentally could lead to injury. On the other hand, a forward dynamics computer simulation model of a gymnast and landing mat could be used to look at the performance enhancement of various landing techniques and gymnastic skills whilst safely assessing the associated injury risk.

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It has been reported that between 57% and 82% of injuries in gymnastics are acute, ranging from strains and sprains to fractures and dislocations, and that the lower extremity was the most injured, comprising 54–70% of all injuries (Andrish, 1985; Harringe et al., 2006; Jensen, 1998; McAuley et al., 1987; Meeusen and Borms, 1992; Snook, 1979). Pettrone and Ricciardelli (1987) found that 31% of acute injuries were fractures to the lower extremity; the foot was mentioned separately, suggesting that the fractures were to the femur and tibia/fibula. Impact forces acting on the human body during landing must be dissipated primarily by the musculoskeletal system and excessive loading may result in injury (McNitt-Gray, 2000). Previous experimental research has focused on landing from various drop heights (DeVita and Skelly, 1992; Dufek and Bates, 1990; McNitt-Gray et al., 1993, 1994; Ozguven and Berme, 1988). The results have shown that the amount of joint flexion, rate of joint flexion, landing phase duration and ground impact forces generally increased with greater drop height. Factors affecting the dissipation of forces during landing have been identified as the landing strategy (Dufek and Bates, 1990; McNitt-Gray et al., 1993) and the landing mat (McNitt-Gray et al., 1994). Landing mats undergo large amounts of area deformation and are now essential for landings from vaults and from gymnastics apparatus dismounts. As large numbers of the injuries in gymnastics occur due to excessive loading of the lower limbs during landings, it is relevant to determine what effect changes in landing conditions have on internal loading.

Within biomechanics both torque-driven models (Alexander, 1990; Chowdhary and Challis, 1999; King et al., 2006) and muscle-driven models (Cole et al., 1996; Denoth, 1985; Spagale et al., 1999) have been used to generate active joint motion. An advantage of torque-driven models is that they can be implemented with subject-specific torque/angle/angular velocity profiles (King et al., 2006). Since torque generators represent the net effect of all the muscles contributing to movement about a joint, they cannot be used to investigate the role of individual muscles. They also fail to account for all the internal forces at the joints, since the action of a linear muscle effects a moment and produces a reaction force in the joint, whereas a torque generator does not.

In a forward dynamics simulation, the joint reaction force (JRF) is calculated as the sum of force vectors at the joint that arise from the mechanical components (for example, rigid bodies, springs and linear actuators) that are present in the model. Simulation time is crucial when attempting to optimise a landing technique, since a large number of iterations are needed. A simpler model may run faster than a more complex model, but it may lack some important elements that are critical to the problems being investigated. As a consequence, it is essential to evaluate any such model with these considerations in mind.

The aims of this study are: (1) to develop a model to investigate whether a subject-specific torque-driven or a subject-specific muscle-driven model of a gymnast (in conjunction with a model of a landing mat) is better at matching measured ground reaction forces (GRFs) and kinematics during gymnastics landings, (2) to calculate their respective simulation run times and (3) to determine what level of model complexity is required to assess injury risk through the calculation of JRFs and bending moments in the femur and tibia within the models. Injury risk will be defined as increased internal loading in terms of JRFs and bone bending moments. The level at which injury will occur will be not be specified as there are no known *in vivo* values for subject-specific injuries. Hence, only the generic risk of injury from increased loading can be assessed rather than the probability of injury.

2. Methods

2.1. Data collection

Three-dimensional motion data, force plate data and electromyography (EMG) recordings were obtained during the landings of one forward rotating (FR) and one backward rotating (BR) vault performed by an elite gymnast (height 1.77 m, weight 75 kg) in a simulated competition environment. The gymnast gave informed consent for all procedures, which were carried out in accordance with the protocol approved by the Ethical Advisory Committee of Loughborough University.

Motion data were recorded at 250 Hz using a 12 camera Vicon 624 motion analysis system in a calibrated volume 3.5 m long $\times$ 2 m wide-3.5 m high with a reconstruction residual of 1.2 mm. Retro-reflective markers (25 mm diameter) were placed at the lateral aspect of the shoulder, elbow, wrist, hip, knee, ankle, and metatarsal–phalangeal joints and the head and toes on both sides of the body. An AMTI force plate measuring 1200 mm $\times$ 600 mm was located beneath a customised landing mat which had a section the size of the force plate separate from the rest of the mat to minimise cross-bridging. Force data were collected at a sampling frequency of 1000 Hz with a 10% pre-trigger and a sampling duration of 5 s. EMG data were collected at 1000 Hz, with a gain of 3000, from the gastrocnemius, tibialis anterior, biceps femoris, vastus lateralis, rectus femoris and gluteus maximus on the on the right side of the gymnast using a portable Biovision EMG system. The EMG system, the force plate and the Vicon motion analysis system were all synchronised using a radio signal.

2.2. Data processing

Personalised segmental inertia parameters for the gymnast were calculated (Table A1) using the inertia model of Yeadon (1990); these were then divided into the rigid and wobbling mass inertial parameters (Table A2) based upon Pain and Challis (2006). Subject-specific strength parameters for the gymnast were calculated (Table A3) using a nine-parameter mathematical function based on the seven-parameter maximum voluntary torque/angular velocity function of Yeadon et al. (2006). The two extra parameters defined the torque/angle relationship and determined the torque/angle/angular velocity relationship using methods similar to King et al. (2006). Obtaining the muscle model parameters involved the subject performing a series of maximal effort isometric trials and eccentric–concentric cycles on a dynamometer.

The Vicon marker positional data were smoothed using a generalised cross-validated spline (Woltring, 1986) and the splined positional data were used to calculate the gymnast's joint angles and trunk orientation angle during the final 150 ms prior to landing and throughout the landing phase of the skill. The splined data were also used to calculate the mass centre velocity, the joint angular velocities and the trunk angular velocity throughout the landings. The raw EMG signals were full-wave rectified