INCORPORATING JOINT COMPLIANCE WITHIN A RIGID BODY SIMULATION MODEL OF DROP JUMPING

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INTRODUCTION

Impact forces of up to 13 times bodyweight have been observed in dynamic jumping activities such as the triple jump [1]. It has long been accepted that the human skeletal system is capable of damping such impact shock waves and avoiding direct transmission of impact forces to internal structures. The force attenuating mechanisms responsible, including foot arch and heel pad compliance; lower extremity joint compression; and spinal compliance, have previously been overlooked in forward-dynamics whole-body simulation models in aid of simplistic representations. Indeed, a general assumption of the existing models has been the simplistic modelling of frictionless pin joints and fixed segment lengths. Pin joint representations have therefore resulted in unrealistic dissipation of force and acceleration throughout the body following impact and hence difficulty in accurately reproducing experimentally measured ground reaction forces [1].

Previous studies have attempted to overcome this limitation by modelling excessive wobbling mass movement or compression at the foot-ground interface to compensate for the lack of compression and thus force dissipation within the joint structures [1,2,3]. Allen et al. [1] stated that whilst unrestricted ground compression was appropriate for simulating performance, accurate internal force replication would require compliance elsewhere within the rigid link system.

The purpose of this study was therefore to investigate the effect of incorporating joint compliance on the ability of a computer simulation model to accurately predict ground reaction forces during dynamic jumping activities.

METHODS

A planar computer simulation model was constructed within AUTOLEV™. The model consisted of nine rigid segments representing the forefoot, triangular rearfoot, shank, thigh, lower trunk, upper trunk, head and neck, upper arm, and lower arm. The model incorporated wobbling mass elements in the shank, thigh and trunk (spanning upper and lower trunk segments). The foot-ground interface was modelled using non-linear spring-damper functions vertically and horizontally at the toe, MTP joint, and heel. The MTP, ankle, knee, hip, shoulder, and elbow joints were each driven by extensor and flexor torque generators, whilst the neck angle did not vary. Ankle plantarflexion and knee and hip flexion and extension were driven by biarticular joint torque generators [4,5] with the joint torque determined by activation level as well as the angle and angular velocity at both the primary and a secondary joint (e.g. knee extension torque determined from knee and hip kinematics).

In addition, viscoelastic elements were incorporated at the ankle, knee, hip, mid-trunk, and shoulder joints connecting the distal end of one rigid segment with the proximal end of the adjacent rigid segment. These represent the internal compliance within the human medio-longitudinal foot arch as well as within the articulating joints and the curvature of the spine. The compliant joint spring-damper force, F_j, was given by

$$F_j = k_j s^3 - \beta_j \dot{s}$$

where k_j and β_j are the stiffness and damping coefficients, respectively, and s and $\dot{s}$ are the stretch and stretch rate of the spring-damper, respectively.

The position of the upper arm insertion along the rigid upper trunk segment was determined by a cubic fit against shoulder joint angle, replicating depression and elevation of the shoulder girdle as the upper arm is lowered or raised respectively.

The simulation model was made specific to a national level male 100 m sprinter (23 years, 1.86 m, 88.6 kg, personal best 10.50 s) using experimentally collected data obtained during drop landings and maximal drop jumps, including arm swing, from drop heights of 0.30, 0.445, 0.595, and 0.74 m. Lightweight Dytran triaxial accelerometers (1000 Hz) were positioned over the first metatarsophalangeal (MTP) joint, the distal and proximal anteromedial aspects of the tibia, the anterolateral distal femur (all on the dominant leg), the L5 vertebra, and the C6 vertebra. The positioning of these accelerometers was measured and accelerations at the same positions on the simulation model were output for the purpose of acceleration attenuation comparison and evaluation.

Rigid and wobbling segmental inertia parameters were determined from anthropometric measurements taken according to the protocol of Yeadon [6]. Subject-specific monoarticular and biarticular joint torque parameters were calculated from maximum voluntary torque measurements at the ankle, knee, hip, and shoulder joints taken using an eccentric-concentric protocol on a Con-Trex isovelocity dynamometer. MTP torque parameters were scaled from those at the ankle.

The stiffness and damping coefficients of the wobbling masses, compliant joint springs, and foot-ground contact springs were determined alongside model evaluation during a matching optimisation process. A parallelised genetic algorithm varied these parameters as well as the torque generator activation parameters to minimise an RMS of cost functions between the model and corresponding experimentally collected whole body kinematic and ground reaction force data for a 0.595 m drop jump, given the same conditions at touchdown.

Penalties were applied to the cost function in any simulation where displacement at a viscoelastic element exceeded predefined anatomical limits. The compliant ankle joint spring was assumed to represent ankle joint compression as well as medio-longitudinal arch depression and navicular drop inferior to this position. Penalty thresholds were determined with reference to the relevant literature. Similarly, displacement limits at the knee and hip were determined with reference to relevant joint space and distraction gap literature. Both the mid-trunk and shoulder spring limits were determined from the collected experimental data of drop jumps and drop landings. The mid-trunk represented the observed resultant length change between the C7 and L5 vertebra, with the shoulder spring replicating the experimental acute change in hip to shoulder distance following impact with the ground.

The novel introduction of compliance within joint structures enabled a reduced magnitude of compliance elsewhere in the system, and thus more realistic displacement limits at the wobbling masses and foot-ground interface. At the foot-ground interface new limits were determined with reference to scientific literature investigating foot-shoe-ground horizontal displacement, and shoe compression with the addition of heel pad compression at the heel. Wobbling mass displacement limits were determined from a spectral analysis of marker movement in relation to the underlying rigid segment during the experimental drop jump and drop landing data collection.

In addition to the above matching and model evaluation process (compliant model), the same process was repeated for comparative purposes with a similar model featuring pin joints in place of the viscoelastic joint springs (rigid model) and the same penalty limits.

RESULTS AND DISCUSSION

The overall difference in kinetic and kinematic time-histories between the compliant model and experimental performance during the evaluation and parameter determination process was less than 5%. This included an RMS of vertical and horizontal ground reaction forces that was also less than 5% of peak vertical ground reaction force. All viscoelastic displacements were within the bounds imposed and so no penalties were incurred. In comparison, RMS differences were greater for the rigid model with traditional pin joints. The difference between model and experimental performance data was less than 10%. Ground reaction forces differed from experimental data by greater than 10%. Again, no penalties were incurred.

Thus, the incorporation of viscoelastic elements at key joints enables replication of experimentally recorded ground reaction forces within realistic whole body kinematics and removes the previous need for excessive compliance at wobbling masses and/or the foot-ground interface. Future research should continue to evaluate the force and acceleration transmission within a compliant model and assess the ability to generate realistic joint reaction forces within a relatively simplistic whole body simulation model.

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